Recovery from a trip in young and older adults

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Recovery from a trip in young and older adults

mechanics and control of the support limb

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ten overstaan van de promotiecommissie
van de faculteit der Bewegingswetenschappen
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De Boelelaan 1105

door

Mirjam Adriana Gijsberta Maria Pijnappels

geboren te Veghel

promotor: prof.dr. J.H. van Dieën

copromotor: dr. M.F. Bobbert

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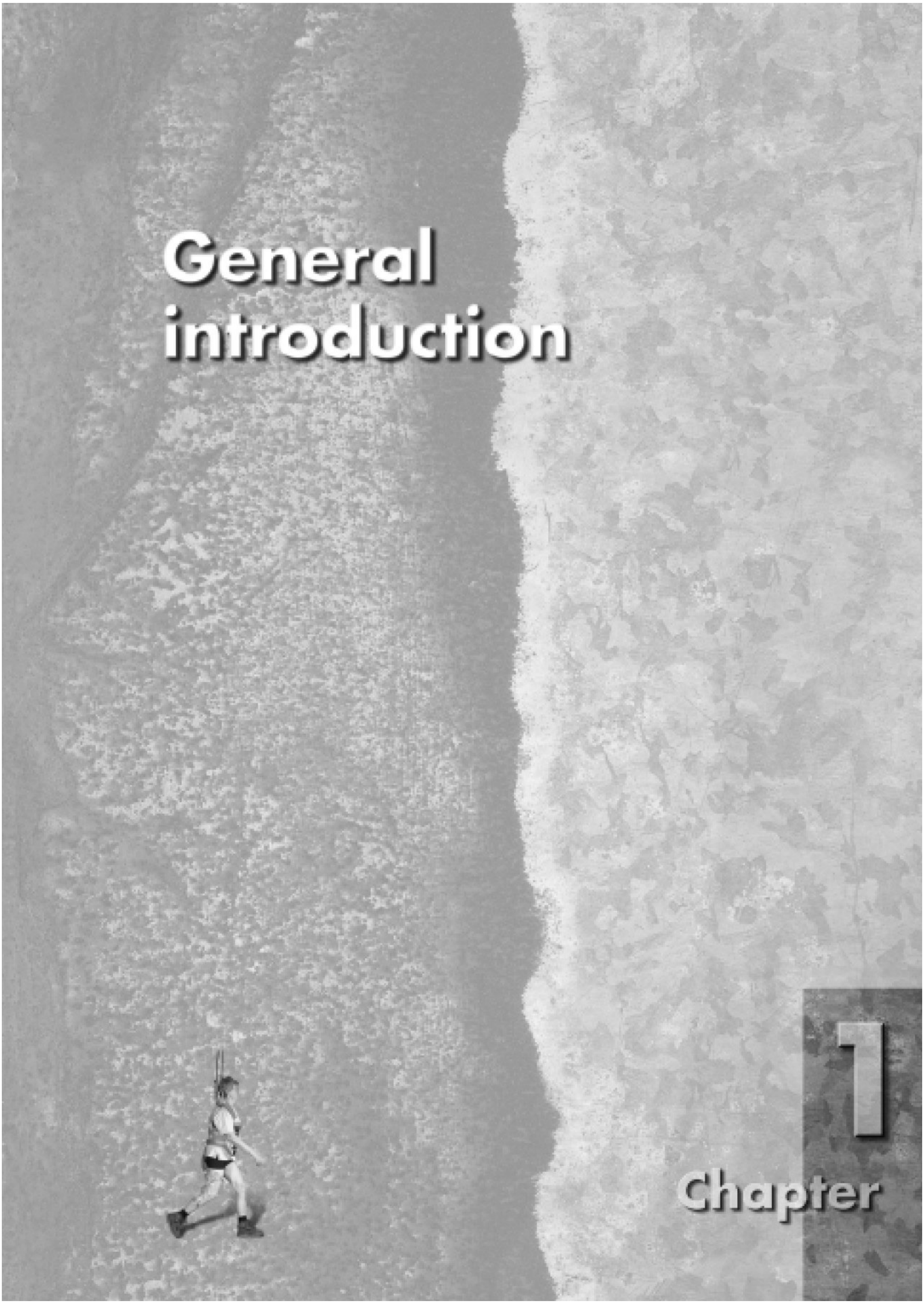

General introduction

1

Chapter

Introduction

Falls and fall-related injuries are the cause of serious medical and social problems, especially in the growing elderly population. In the search for intervention targets to prevent falls and their consequences, identification of the risk factors and causes of falls is needed. Tripping is one of the main causes for falls in older people. This thesis focuses on the mechanics and control of recovery reaction after tripping. Its main purpose is to obtain insight into the requirements for a successful recovery reaction after tripping and to understand why older people sometimes fail to meet these requirements. This might help to identify causes for inadequate reactions and falls and to identify targets for prevention.

In this general introduction, first an overview will be given of the epidemiological studies on falls in the elderly to illustrate the importance of the problem. Then, the factors determining the probability of a fall as a consequence of a trip will be described. Furthermore, the knowledge thus far and the questions remaining on the characteristics of recovery reactions after tripping will be defined. Finally, the aims and the outline of this thesis will be specified.

Risk factors for falls in the elderly

Among the community-dwelling elderly people, 25% of persons over 65 years and 35% of persons over 75 years experience at least one fall per year [91]. About 50% of elderly persons who fall, experience multiple falls within one year. In institutional settings like nursing homes, the frequency of falls is considerably higher than in the community, with 50% of the people experiencing a fall at least once per year.

Although most falls do not cause serious injury, major injuries affect around 5% of the elderly each year. Among these are fractures (mainly of hip and distal forearm), head trauma, musculoskeletal injuries, and serious soft tissue injuries. Furthermore, up to 90% of elderly fallers have been reported to suffer consequences such as restricted activity, immobility, and adverse psychological effects [2, 43, 79, 91, 93]. Falls often lead to admission into a nursing home [94] and are a leading cause of accidental death in senior citizens [68]. Estimates of the mortality rate in older persons one year after a hip fracture, which in 90 % of the cases is the result of a fall, vary from 14-36% [91], to 50% [43].

Most of the falls experienced by older people take place during commonplace everyday activities like walking, ascending or descending stairs, and transferring on or off chairs and beds [2, 44, 70, 91, 92]. Trips and slips are the most common causes of falls, accounting for 17% up to 60% of falls [2, 25, 44, 70].

Identification of the risk factors and causes of falls can provide intervention targets in fall prevention programs [6]. In the literature, an abundance of observational studies has been published on predictors of falls in older people. An extensive overview of the literature was published by Lord et al. [36]; the main findings were summarized by van Dieën et al. [98]. In short, predictors for falls comprise intrinsic factors (e.g. psychosocial and demographic factors, postural instability, sensory and neuromuscular factors, medical factors and medication use) and environmental factors (i.e. factors that increase the probability of falls, like the presence of obstacles, slippery surfaces, curbs and stairs, poor lighting and unsuitable footwear). Most falls result from an interaction between intrinsic and environmental factors and, not surprisingly, the probability of a fall increases when multiple risk factors are present [8, 43, 81, 91, 93].

Unfortunately, many of the factors identified as predictors for falls in epidemiological studies (e.g. female gender, depression, incontinence, or hand grip strength) cannot be considered risk factors for falls, because there is no biologically plausible causal mechanism explaining the association [29]. Therefore, these mostly indirect associations do not give adequate directions for preventive measures. For example, a significant correlation between recurrent falling and the intrinsic predictor hand grip strength [82] does obviously not imply that training of grip strength is indicated to prevent falling. Hand grip strength might be associated with lower extremity strength, which could be a limiting factor in balance recovery, although the predictive value of knee extensor strength was not better than that of grip strength [82].

Whereas epidemiological studies are limited to identification of risk factors for falls, experimental studies on falls can provide more detailed insight into causal factors [6, 98]. Hence, current studies are focused on responses to mechanical perturbations, for example responses to experimentally induced tripping [20, 22, 27, 50, 72, 78] and slipping [9, 38, 65, 85]. In line with these studies, the present thesis describes experiments in which both young and older subjects are tripped and challenged to recover without falling.

The probability of falling after a trip

It can be questioned whether older people fall more often than young people because they trip more often or because they are less able to regain balance after a trip. The probability of tripping depends on the presence of obstacles and on the individual's walking pattern. For instance, age-related gait changes in the walking pattern (e.g. reduced speed, stride length and toe clearance and increased double support time) might enhance stability, but might also increase the probability of tripping (for an overview see [98]). Studies on obstacle avoidance strategies suggest that age-related changes (e.g. reduced toe clearance, increased reaction times, and visual and cognitive impairments) indeed negatively affect obstacle avoidance success [10, 11, 40, 48, 101]. However, these studies could not confirm that older subjects actually trip more often than young subjects.

In a large survey based on community-dwelling individuals over 65 years of age [4], tripping caused 57.5% of the falls in those aged 65-69 years, but this prevalence decreased to 29.7% in those aged 85 years and over. This may be a reflection of a higher probability of tripping due to the greater mobility of the young and strong elderly compared to the old and frail elderly people. Moreover, in an experimental study, in which older subjects were tripped [49-52], it was also found that the relatively young and strong people in this elderly group had an increased likelihood of falling after a trip. Strong older people obviously walk faster, which makes recovery after the trip more demanding. In the abovementioned experiments, 10 out of 61 of the older subjects actually fell after tripping and substantial forces were recorded in the safety rope of another 12 subjects (who had probably fallen had there been no safety rope). In contrast, in two earlier studies of the same group on young adults, all subjects were able to recover successfully [26, 27]. In another experiment on both young and older subjects, a trip was induced during treadmill walking [23]. In these experiments, only one fall occurred over a total of 36 trips in four young adults, whereas six falls occurred over a total of 19 trials when two older adults were tested.

So, the probability of tripping might be related to age-related changes in the walking pattern and the fitness of the individual, but the question remains whether older people trip more often than young people. However, the probability of recovering successfully from a trip is clearly lower in elderly subjects than in young adults. Hence, this thesis will focus on the requirements for a successful recovery reaction after a trip and to understand why older people sometimes fail to meet these requirements.

Recovery reactions after tripping

In a trip, impact with an obstacle induces a forward rotation of the body. Without an appropriate response of the subject, this rotation is further accelerated by gravity and the body will fall on the floor. Early experimental studies on tripping, which used only minor perturbations of the swing leg of a subject walking on a treadmill, showed that responses of the leg muscles depend on the time of the swing phase at which the perturbation occurs [3, 16]. Eng et al. [20] discerned two strategies for recovery when young subjects were tripped over an actual obstacle. Which of these two occurs depends on the time of trip initiation in the swing phase (Figure 1.1).

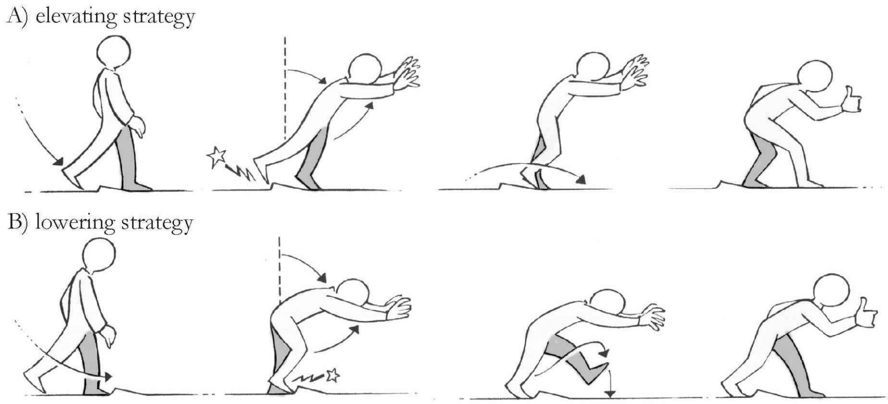

Figure 1.1: *Recovery strategies after tripping (*Stilstaan bij bewegen © *Natuur & Techniek, 2001).*

An elevating strategy can be observed after a perturbation in early swing and consists of an elevation of the obstructed (ipsilateral) swing limb to overtake the obstacle. A lowering strategy is seen after perturbation in late swing and consists of an immediate placement of the obstructed foot on the ground, followed by a next step to overtake the obstacle. For both strategies, the foot that is positioned forward after the trip is coined the recovery foot. Note that in the elevating strategy, the recovery limb is the limb that was obstructed during the trip, whereas in the lowering strategy it is the limb that was in stance at trip initiation. Schillings et al. [74] further investigated these strategies during tripping on a treadmill and described a transition from one strategy to the other around mid-swing.

Grabiner et al. [27] described the reactions after tripping as consisting of two phases. The phase from impact with the obstacle until placement of the recovery foot was coined positioning phase. Placement of the recovery foot initiates what was called the recovery phase. The term for the primary phase, positioning phase, suggests that the essence of this phase is to position the recovery limb through reactions in this limb. Indeed, when the recovery limb is placed properly, i.e. anterior of the body center of mass, it can generate a moment that counteracts the body's forward rotation [27]. However, other actions can also contribute substantially to an adequate recovery during this primary phase. While the recovery limb is being positioned, but before it hits the ground, a strong push-off reaction can be generated in the support limb. Given that the literature on tripping thus far focused on the swing limb, little is known about the exact role of the support limb in recovery after tripping. Theoretically, adequate force generation during push-off by the support limb can reduce the angular momentum of the body. The more reduction is achieved by the support limb, the less remains to be accomplished by the recovery limb. Whether and how push-off by the support limb contributes to recovery has never been investigated in young or in older subjects.

Generating rapid and strong push-off reactions by the support limb could be a problem for the elderly people, since lower extremity strength and the rate of force generation are known to decline with age [36, 75]. Declined force generation can be due to changes in muscle properties or to changes in neural control [63]. Successful recovery requires rapid selection and execution of an adequate response, which might be too demanding for older adults because of age-related changes in latencies, sequencing, and amplitudes of functional responses to postural perturbations [25, 83, 106].

Aims and outline of this thesis

The aims of this thesis were to obtain insight into the requirements for a successful recovery reaction after tripping - in particular the mechanics and control of the support limb - and to understand why older people sometimes fail to meet these requirements. Series of tripping experiments have been conducted on both young and older subjects, in order to obtain the required information. The chapters arising from these studies were on the following topics: ecological validity of the experimental setup, the role of the support limb in successful recovery, and age-related changes in mechanics and control of the support limb reactions.

First, an experimental setup was developed in which subjects could be tripped repeatedly during over-ground walking. As the subjects had to be informed about the purpose of the study for ethical reasons, changes in the walking pattern might occur. Such changes could cause recovery reactions to be different from recovery reactions in real life. In Chapter 2, the question was addressed whether forewarning of a possible trip changes the walking pattern in young subjects in terms of kinematics and kinetics. Normal walking patterns, collected at the beginning of the experiment when subjects were assured that they would not be tripped, were compared in kinematic terms with patterns of forewarned walking, collected in between actual tripping trials. In Chapter 3, a similar comparison was made for muscle activity patterns in young as well as in older subjects.

Secondly, the contribution of the support limb to recover from a trip was investigated in young subjects. The support limb was hypothesized to contribute in two ways: (a) by providing time and clearance for proper positioning of the recovery limb, and (b) by restraining or reducing the forward angular momentum of the body induced by the trip. In Chapter 4, a method was developed to calculate the angular momentum and the two hypothesized ways of support limb contribution to recovery after tripping were investigated for the elevating strategy. In Chapter 5, the question was posed of how this contribution to recovery is achieved during push-off by the support limb in terms of muscle activity and moment generation.

Thirdly, limitations of older subjects in the mechanics and control of the support limb recovery reactions after tripping have been identified. In Chapter 6, it was questioned whether older subjects react less adequate than young subjects during recovery after tripping. Support limb moment generation (in terms of onsets, rates of development and peak values) were compared between young subjects, older non-fallers, and older fallers. In Chapter 7, it was investigated whether control of muscle responses after tripping differs between young and older subjects. Differences in timing and sequencing of muscle activation, as well as differences in the magnitude and rate of development of muscle activation were addressed.

Finally, in Chapter 8, the main findings and conclusions of this thesis were summarized and a number of recommendations for further research and for fall prevention were provided.

Changes in walking pattern caused by the possibility of a tripping reaction

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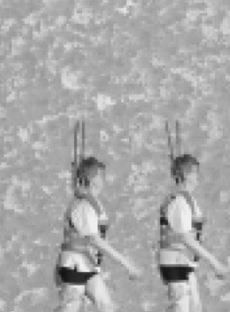


2

Chapter

Abstract

This study investigated in 15 young adults whether the walking pattern was altered after forewarning of a possible trip. Such changes might affect tripping reactions and consequently the validity of experimental results. Kinematics and dynamics were measured during over-ground walking. No changes occurred in walking velocity, step frequency, duration of stride cycle, stance, swing and double support time, and step length. A small increase was found in step width and foot clearance (due to ankle dorsiflexion), but these changes are not expected to alter the probability of tripping nor the recovery reactions after tripping in an experimental setup.

Introduction

The population of elderly people in western society has increased considerably the last decades. Consequently, the effects of aging on medical and social health have increased too. Falls and fall-related injuries are common, costly and serious medical problems for the elderly. One in three adults over 65 years of age falls once a year, mostly as the result of a trip or slip [2, 44, 70]. In order to reduce the occurrence of trip-related falls, identification of the factors that increase an individual's risk of falling following a trip is needed.

For investigation of recovery reactions from mechanical disturbances during locomotion in old and/or young adults, an experimental setup is required in which unexpected tripping can be provoked [107]. Several studies have investigated tripping in a laboratory setting. Schillings et al. [72-74] investigated reflex responses during stumbling over obstacles in both young and old adults while walking on a treadmill. Grabiner et al. [27] and Eng et al. [20] let young adults trip repeatedly over suddenly appearing obstacles during overground walking. Pavol et al. [46, 49, 50] used a similar method of obstacles unexpectedly appearing from the ground. They were able to let older adults trip, although in these tests the experiments were limited to a single trip attempt.

In all these studies, recovery reactions were compared with normal walking. For ethical reasons, subjects were informed in advance about the purpose of the study. This raises the question of whether (and how) subjects change their walking pattern when they are forewarned for a possible trip. From studies on obstacle avoidance it is known that subjects may reduce their walking velocity, step length and increase foot clearance [1, 10, 12, 39]. If these changes in the walking pattern also occur under postural threat in a laboratory setting, resulting recovery reactions may differ from recovery in real life.

Aforementioned authors all refer to the possibility of anticipatory behaviour. The comparisons of walking pattern between normal and test walking were limited to the parameters walking velocity [27], stride length [50] and a subjective description of step cycle duration, joint angle and EMG activity in one subject [72]. None of them reported any differences with normal walking.

In contrast, Eng et al. [20] did observe anticipatory behaviour, which appeared not to be based on the release of the obstacle, but rather on the threat of the task. They tried to minimize it by a large number of "catch" trials. Moreover, trials (and subjects) in which EMG signals demonstrated anticipatory behaviour were excluded. Unfortunately, Eng et al. [20] did not describe how anticipatory behaviour was expressed.

In the present study, an experimental setup is tested to let subjects trip. Our main concern is that anticipatory or adaptive changes in the walking pattern will occur, which might affect the probability of tripping and/or the recovery reaction. The goal of this study is therefore to investigate whether forewarning of tripping leads to such changes in the walking pattern of young adults. Variables under consideration are firstly general parameters, subdivided in temporal and spatial components and secondly foot elevation parameters. Besides calculation of kinematic data, ground reaction forces are analyzed. Averaged magnitude as well as variability of the variables will be considered.

Methods

Participants

Fifteen volunteers (8 male, 7 female) participated in this study. Mean age (± SD) was 27 (± 4) years, mean height 1.83 (± 0.10) m and mean weight 76 (± 10) kg. Subjects were informed on the research procedures before they gave informed consent in accordance with the ethical standards of the declaration of Helsinki.

Protocol

The experimental procedures used in this study have been approved by a local ethical committee. Subjects, wearing walking shoes, were instructed to walk at a self-selected velocity over a platform of 10 meters in which a force plate was mounted after 5 m (Figure 2.1). Before testing, they were allowed to become accustomed to the testing environment, which involved wearing a full-body safety (parachute) harness and modified glasses.

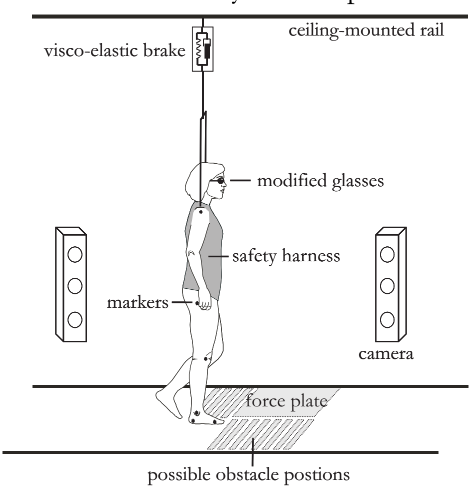


Figure 2.1: *Schematic diagram of the experimental setup from a side view.*

The safety harness, attached to a ceiling-mounted rail, ensured that subjects would not become injured should their recovery reaction be inadequate. Modified glasses, blocking the lower half of the visual field, allowed subjects to look in the walking direction, but prevented them from seeing the platform on which they were walking and the obstacle. In the platform, a wooden obstacle of 15-cm height (40-cm width) could be placed at different positions on either the left or the right side of the walkway (Figure 2.1). During 10 trials at the beginning and 5 trials at the end of an experimental session, subjects were assured that they would not be tripped so that the normal walking pattern could be recorded. After the first 10 trials, subjects were forewarned that a trip might be induced during walking. At the start of each trial, subjects did not know whether an obstacle was positioned, and if any, where. In about 10 of 50 forewarned trials, an obstacle was actually positioned to let the subjects trip.
Gait kinematics were recorded during each experimental trial using 4 Optotrak cameras operating at 100 Hz. Motion was tracked of 12 infrared-light emitting markers, which were placed bilaterally on the anatomical landmarks heel, metatarsol-phalangeal joint (MTP5), lateral malleolus, estimated knee center, trochantor major of the femur, and acromial process. Furthermore, ground reactions forces of the left foot were recorded by a custom-made strain gauge force plate (1x1m) at a sample frequency of 1000 Hz.

Data analysis

First, a few trials were discarded because either kinematic of dynamic data were incomplete. For each subject 5 trials of normal walking and 5 trials of forewarned walking were randomly selected from successful trials. Heel strike (HS) and toe-off (TO) had to be detected on the basis of kinematic data, as force plate data were not available for the right foot. Algorithms for this purpose were recently proposed by Hreljac and Marshall [30]. Unfortunately, they did not produce accurate detection for our data. Therefore, using trials in which both ground reaction forces and kinematic data were available, we developed an alternative way to detect event times of heel strike and toe-off in our data. Time of HS correlated closely to the time of a local minimum in the vertical velocity component of the toe marker. Time of TO was closely correlated to the time of a local maximum in the vertical velocity component of the heel marker.

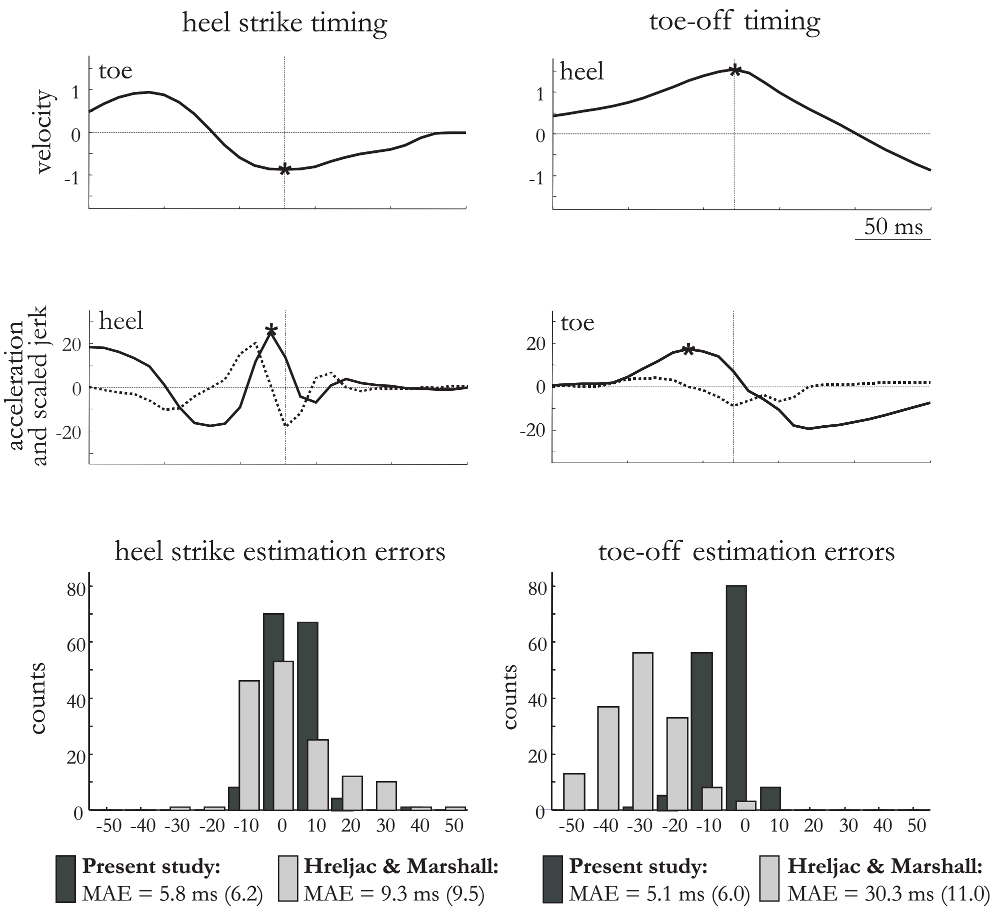


Figure 2.2: *Detection of heel strike (HS; left graphs) and toe-off (TO; right graphs) on the basis of kinematic data using two different methods. Upper two windows show detection of HS and TO according to the method of the present study, in which HS was estimated at the minimum in vertical velocity of the toe marker, and TO estimated at the maximum in vertical velocity of the heel marker. The middle two graphs illustrate Hreljac and Marshall's method. They estimated HS at the local maximum in vertical acceleration (zero jerk; dotted line) of the heel marker and TO at the local maximum in vertical acceleration of the toe marker. Estimations of event timing based on kinematic data are indicated by asterisks, and timing of HS and TO based on force plate data are indicated by vertical dashed lines. The lower two histograms illustrate the frequency distribution of true estimation errors (negative for an early prediction) in 150 trials for heel strike and toe-off event timing according to both detection methods. Mean absolute error (MAE) and SD of the two methods are represented for each timing event.*

Figure 2.2 represents estimation of event timing in a typical example according to our method (upper two graphs) and to that of Hreljac and Marshall [30] (middle two graphs). Estimations of timing events of the left foot were compared with HS and TO based on synchronized force plate data (rise of vertical component of the ground reaction force above 25 N) and resulted in an overall mean absolute error (MAE) of 5.5 (± 6.1) ms for our method, whereas the MAE was 19.8 (± 14.7) ms when events were based on the algorithm of Hreljac and Marshall [30] (lower two bar plots). MAE as well as error range were smaller according to our method for HS and particularly for TO. Apart from smaller MAE and error range, determining local maxima and minima in vertical velocity was easier to apply than finding maxima in the more fluctuating acceleration signal.

Based on HS and TO events, the parameters of the variables under consideration were analyzed. To get mean results for both normal and forewarned walking conditions for each of the calculated parameters, values were averaged over left and right side, trials and subjects. Variables under consideration were first of all general variables, subdivided in temporal components (i.e., velocity, step frequency, stride cycle time, stance time, swing time and double support time), and spatial components (i.e., stride length, step width or lateral heel distance and minimum foot clearance during swing phase). Foot clearance (minimum toe height) during swing was calculated as the difference in vertical position of the MTP5 markers with respect to the contralateral (stance) leg, corrected for differences in marker placement. At the instant of minimum toe height, foot elevation variables were additionally calculated (i.e., timing, segment and joint angles at minimum toe height). Although the probability of tripping was expected to be highest at minimum toe height, time series of toe position and joint angles over the entire stride cycle were also considered. Furthermore, time series of ground reaction forces were analyzed to describe dynamical results. Time series were normalized to 100% of total stride time. Finally, variability over stride cycles within subjects was calculated over time series and expressed as the coefficient of variation (CV), according to Winter [102].

Statistical analysis

In order to test for a difference between normal and forewarned walking patterns, within-subject averaged (across trials) values of both temporal and spatial parameters were analyzed in a multivariate analysis of variance (MANOVA) for repeated measures. An additional MANOVA was performed on the foot elevation parameters timing, segment and joint angles at minimum toe height. Preliminary analysis including the factor *trial* revealed no significant main effects, nor interactions involving this factor. Therefore, all further reported results are averaged over trials. The level of significance selected for this study was $p<0.05$.

Results

Mean values over sides, trials and subjects and standard deviations, as well as results of testing differences in the considered parameters between normal and forewarned walking pattern, are presented in Table 2.1. Multivariate analysis revealed a significant overall effect of forewarning of a possible trip on the walking pattern. Specified univariately, this effect was attributable to significant increases of step width (1.2 cm) and minimum toe height (1.1 cm). Thus, none of the temporal parameters was altered after forewarning, whereas the spatial parameters were increased, except for stride length. The effect of forewarning on foot trajectory is illustrated in Figure 2.3, depicting the trajectory of the averaged position of MTP5 over a normalized stride cycle for the two walking conditions.

Table 2.1: *Mean and standard deviation (SD) values of temporal and spatial parameters for normal and forewarned walking pattern; n.s. = not significant.*

	Normal	Forewarned	Significance
Temporal			
velocity (m/s)	1.58 (0.15)	1.57 (0.19)	n.s.
frequency (steps/min)	113.66 (7.20)	114.70 (7.09)	n.s.
cycle time (s)	1.06 (0.07)	1.05 (0.06)	n.s.
stance phase(% cycle time)	58.54 (1.27)	58.59 (1.13)	n.s.
swing phase(% cycle time)	41.47 (1.29)	41.41 (1.13)	n.s.
double support phase (% cycle time)	8.63 (1.31)	8.86 (1.09)	n.s.
Spatial			
stride length (m)	1.67 (0.15)	1.65 (0.17)	n.s.
step width; lateral heel distance (cm)	23.27 (2.66)	24.50 (2.44)	$F(1,14) = 7.3$ $p<0.05$
min. toe height during swing (cm)	2.19 (0.66)	3.32 (1.08)	$F(1,14)=31.2$ $p<0.00$

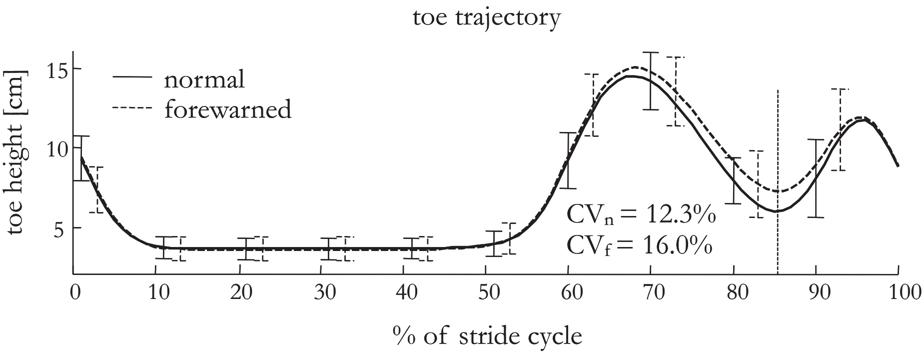


Figure 2.3: *Averaged (± SD) and normalized trajectory of toe position during a stride cycle for normal (solid line) and forewarned (dotted line) walking. The vertical dashed line represents minimum toe height (on average at 85.4% of the stride cycle). CV=coefficient of variation; n=normal walking; f=forewarned walking.*

The additional multivariate analysis on foot elevation parameters at minimum foot clearance resulted in a significant difference. According to the univariate tests, relative timing of minimum toe height in the stride cycle was not different between the walking conditions. Elevation of the toe after forewarning at minimum height could not be attributed to differences for hip or knee angle, but to a significant increase of dorsiflexion for ankle joint angle at the time of minimum toe height (2.6°). In fact, the increase of ankle dorsal flexion was seen throughout the entire swing phase of forewarned walking (Figure 2.4), whereas the difference in toe position peaked at minimum height. Nevertheless, the foot segment was about 180° at mid-swing in both walking conditions. Hip and knee joints angles did not display any large differences over the stride cycle (Figure 2.4). The increased dorsal flexion in the ankle joint after forewarning corresponded with significant increases in foot and shank segment angles of respectively 3.8° and 1.2°, whereas thigh and trunk segment angles did not differ between walking conditions. Furthermore, curves of ground reaction forces did not show any noticeable differences in dynamics between normal and warned walking (Figure 2.5). Finally, calculation of the coefficient of variation over the stride cycle resulted in larger CV's for all parameters during forewarned walking, indicating larger variability in the walking pattern for this condition (Figures 2.3, 2.4 and 2.5).

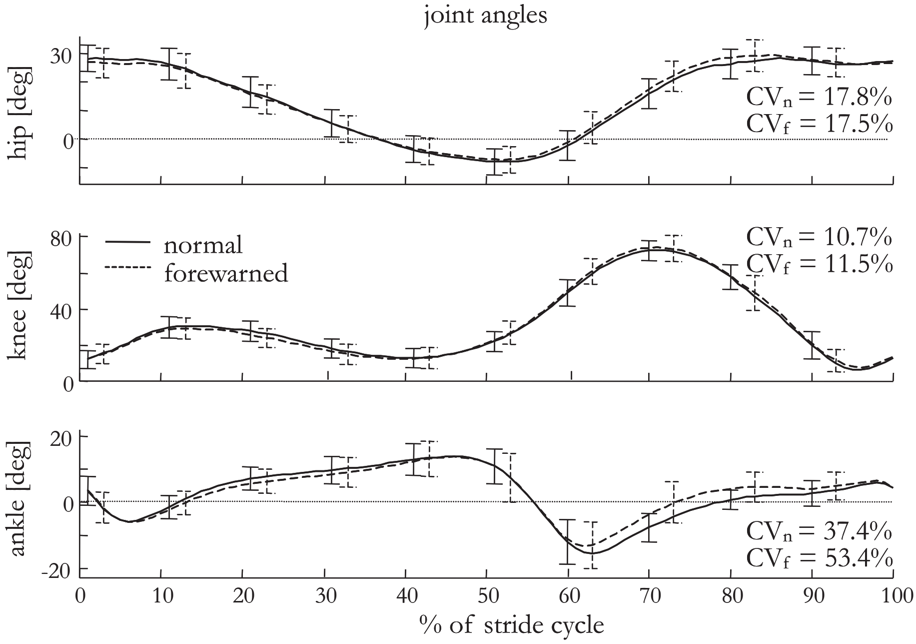


Figure 2.4: *Averaged (± SD) trajectory of joint angles (hip, knee and ankle) during a stride cycle for normal (solid line) and forewarned (dotted line) walking. Positive ankle joint angle=dorsiflexion; positive knee joint angle=flexion; positive hip joint angle=flexion; CV=coefficient of variation; n=normal walking; f=forewarned walking.*

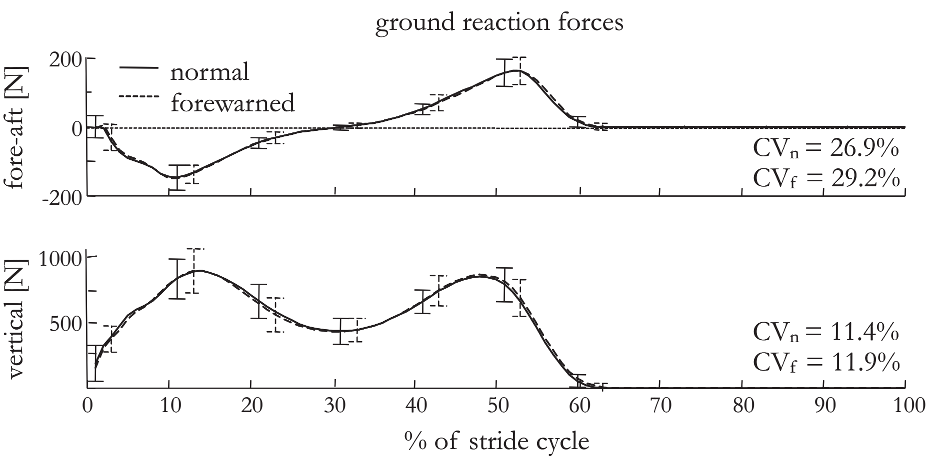


Figure 2.5: *Averaged (± SD) trajectory of fore-aft and vertical ground reaction forces during a stride cycle of the left foot for normal (solid line) and forewarned (dotted line) walking.*

Discussion

Changes in the walking pattern after forewarning

This study investigated whether changes occurred in the walking pattern after forewarning of a possible trip. Compared to normal walking, the pattern after forewarning was not altered in terms of temporal parameters. Of the spatial aspects, step length was not changed, but step width and foot clearance were increased by forewarning. Relative widening of steps was small (5.3%). The relative change of toe elevation, on the other hand, was larger (51.6%) and was attributed mainly to a more dorsiflexed ankle joint throughout the whole swing phase. As expected, the largest difference in toe elevation between normal and forewarned walking occurred at the instant of minimum toe height. A difference in the ankle joint was not only seen for absolute joint angle values, but also for the variability (i.e., CV) of the joint angles. The variability for ankle joint angle was clearly increased during forewarned walking, especially during swing phase. Although CVs of other calculated parameters did not result in such a pronounced difference, they were all larger after forewarning, except for hip joint angle. Therefore, the forewarned walking condition can be described as less consistent than normal walking, possibly because subjects adopted a more secure, but less accustomed walking pattern. Despite some changes in kinematic data between the walking conditions, the trajectories of dynamics did not show any large differences – even though forces are related to accelerations and therefore can be considered as derivatives of position data.

Finally it should be noted that the presented results are averaged data over 15 subjects. Of course, each individual has a specific walking pattern and might exhibit different (degrees of) alterations in the walking pattern. For example, minimum foot clearances during swing ranged from 0.1 to 7.2 cm between trials and subjects. In concern of experiments in which mean results over a group are described, like the present study, these individual differences are not of specific relevance. However, for specific case studies, consideration of the individual's changes in walking pattern is recommended.

Consequences of changes on tripping experiments

The purpose of the present study was to investigate whether walking pattern is altered in young adults after forewarning of tripping. The motivation was that such anticipatory or adaptive changes might affect the probability of tripping and/or the recovery reaction, resulting in tripping reactions in the experimental setup dissimilar to those in real life. To understand the consequences of the observed changes in walking pattern on either the risk of tripping or the recovery reactions, knowledge concerning these matters derived from earlier studies on obstacle avoidance and tripping has to be considered.

In obstacle avoidance, strategies are required to minimize the risk of interference with the obstacle. Foot-obstacle clearance is increased in comparison with normal foot-floor clearance - even when the obstacle is no more than a tape on the floor [1, 10, 12, 13, 39]. Heel or midsole most frequently is the lowest point of the shoe at crossing the obstacle, which may carry less risk for a forward fall in case of contact than when the toe first contacts the obstacle [10]. Furthermore, elevation of the feet to clear the obstacle during avoidance was mainly attributable to increased knee flexion [1, 12, 13, 39]. In the present experiment, feet were also elevated higher. Although ankle joint and foot segment angle were slightly increased over the entire swing phase, the toe remained the lowest point of the foot up to minimum toe height, just as during normal walking. Moreover, in contrast to changes in the ankle joint, knee and hip joint angles did not show any systematic strategies. So, although the feet are elevated approximately one centimeter higher because of ankle dorsiflexion, the toe remains the lowest point at mid-swing. Therefore, we assume the chance on hitting the obstacle of 15-cm height with the toe remained the same in our experiment.

Studies on tripping and recovery reactions provide information on parameters that influence the probability of falling. Increased walking speed and stride length, for example, are described to increase the likelihood of falling following a trip [49]. On the other hand, step width, trunk flexion and phase of gait in which the trip occurred, did not affect the likelihood of falling [49]. During recovery from a trip, forward rotation of the body (which depends on walking velocity) must be arrested. Trunk, hip and knee flexion angles play a very important role in this [27]. The perturbed leg is either placed down in front of the obstacle or lifted over the obstacle, dependent on the phase of gait in which the trip occurs [20, 74]. Both strategies require collaboration of the lower limb joints of both swing and the stance limb and control of the center of mass. Recapitulating, walking velocity, stride length, trunk, hip and knee flexion angles are thus described as determinants in tripping and recovery reactions. None of these parameters were altered in the present study after forewarning. The observed change in step width is not expected to have effect on the recovery reaction. Only the increased dorsal flexion in the ankle joint during swing phase might be of concern during the recovery response strategies.

Finally, the higher variability of the walking pattern after forewarning of a possible trip should not be ignored. Impaired control of foot trajectory during the swing phase of gait has recently been hypothesized to increase the risk of slipping or tripping, but the predictive validity of foot trajectory measures with respect to incidence of falls has not yet been established [31].

Conclusion

Alerting for a possible trip in an experimental setup leads to a change in the walking pattern of young adults in terms of several spatial parameters, but not in terms of temporal parameters. The changes are small, however, and are not expected to alter the chance on tripping nor the recovery reactions after mechanical disturbance during walking in an experimental setup.

EMG modulation in anticipation of a possible trip during walking in young and older subjects

Journal of Electromyography and Kinesiology, conditionally accepted

Chapter 3

Abstract

This study investigated whether muscle activity patterns during walking are altered after forewarning of a possible trip in 11 young and 11 older subjects. Changes in muscle activity patterns could affect tripping responses and consequently the ecological validity of experimental results. Electromyograms were measured during normal walking and during walking after forewarning of a possible trip. The area under the EMG curve was calculated. After forewarning, statistically significant increases were observed in the muscle activity patterns of hamstring, quadriceps and dorsiflexion muscles. Generally, the effects of forewarning on the EMG patterns were the same for both age groups, but the patterns of older subjects did not result in significant differences, probably due to their increased variability in muscle activity. Especially, an increased activity was seen in the TA muscle of the left limb that was perturbed most often. This is in line with findings of small increases in the minimum toe clearance during mid-swing. The magnitudes of all increased activities were small compared to the magnitudes of tripping responses reported in the literature. Therefore, anticipatory activity is not expected to have substantial effects on muscle responses in experimental measurements on tripping in neither young, nor older subjects.

Introduction

Falls and fall-related injuries cause serious medical problems, especially in elderly people. A lot of research has been done recently on identification of fallers and on fall-prevention (for an overview, see [6]). A current trend in research on falls is the investigation of recovery reactions after a mechanical perturbation during locomotion, e.g. tripping [20, 23, 27, 50, 72, 74, 78], slipping [9, 38, 65, 85], or waist pulls [42]. These investigations can provide insight into the factors that determine the success of a recovery reaction and might lead to identification of the causes of an individual's risk on falling.

In the aforementioned studies, recovery reactions elicited by mechanical perturbations were compared with normal walking patterns. As the subjects were informed in advance about the purpose of the study for ethical reasons, changes in the walking pattern might appear. Such changes could affect recovery reactions and consequently the validity of experimental results, as resulting recovery reactions may differ from recovery in real life.

In a previous study, we questioned whether and how young adults changed their walking pattern when they had been forewarned for a possible trip [54]. Kinematics and dynamics were measured during overground walking. After forewarning, no changes were found in walking velocity, step frequency, duration of stride cycle, stance, swing and double support time, or step length. Only a small increase was found in step width (1.2 cm) and minimum foot clearance (1.1 cm). The latter was due to increased dorsiflexion in the ankle. These changes were not expected to alter the probability of tripping, nor the recovery reactions after tripping in an experimental setup.

From the fact that kinematic patterns were only minimally affected it must be concluded that net joint moment patterns were only minimally changed. However, it is theoretically possible that the same net moments were produced with more co-contraction, leading to more stiffness and damping in case of perturbations. Although Schillings et al. [72] did not observe changes subjectively in the muscle activity in one subject, Eng et al. [20] did report observation of anticipatory behavior, triggered by the threat of a possible trip. They tried to minimize anticipation by a large number of "catch" trials, and excluded trials (and subjects) in which muscle activity signals demonstrated altered activity prior to the perturbation. Unfortunately, these authors did not describe how the muscle activity was altered and whether specific muscles were particularly affected.

Studies on muscle responses after perturbations mainly focus on young adults, but because such studies on falls focus on groups with a high risk of falling, similar perturbation experiments can and will be performed on older subjects (e.g. [85]). Moreover, it is not unlikely that the muscle activity pattern of older subjects differs from that of young subjects and that possible anticipatory activity is expressed differently.

The present study was designed to complement our former study and investigated whether adaptive changes in the muscle activity patterns of lower limb muscles occurred after forewarning of a possible trip in both young and older subjects. Subjects walked several times over a platform, while the electromyograms (EMG) of lower limb muscles were measured. We compared the EMG patterns of normal walking, collected at the beginning of the experiment when subjects were assured that they would not be tripped, with the EMG patterns of forewarned walking, collected in between actual tripping trials.

Methods

Participants

Eleven young and eleven older subjects voluntarily participated in this study (Table 3.1). Subjects were informed on the research procedures before they gave informed consent, in accordance with the ethical standards of the declaration of Helsinki.

Table 3.1: *Subject characteristics; group averages (and SD).*

Group	# Subjects	Gender	Age (yr)	Height (m)	Weight (kg)	Velocity (m/s)
young adults	11	5 ♂, 6 ♀	27.3 (4.5)	1.78 (0.07)	74. 8 (9.3)	1.61 (0.15)
older adults	11	4 ♂, 7 ♀	67.6 (2.7)	1.72 (0.11)	77.0 (9.6)	1.44 (0.18)

Experimental setup and protocol

The protocol was the same as previously described [54]. The experimental setup was similar to the previous study, but further developed for better control of tripping trials. Subjects walked, without visual field restrictions, at a self-selected speed over a 12 by 2.5 m platform. In the platform, a force plate was mounted and 21 aluminum obstacles (15 cm height) were hidden over a total distance of 1.5 m. During 10 trials at the beginning of an experimental session, subjects were assured that they would not be tripped so that the normal walking pattern could be recorded. After these first 10 trials, subjects were forewarned that a trip might be induced during walking. At the start of each trial, subjects did not know whether an obstacle would appear. Online kinematic data was used to calculate where and when an obstacle had to appear to initiate a trip at mid-swing. In about 10 of 50 forewarned trials, the subjects were actually tripped. Subjects wore a full-body safety harness, attached to a ceiling-mounted rail.

Data collection and analysis

Twelve infrared-light emitting markers were bilaterally placed on joints, to define 7 body segments. The markers were tracked using 4 Optotrak cameras (Northern Digital). Ground reactions forces were measured with a custom-made strain gauge force plate (1x1 m). The kinematic data and ground reaction forces were collected synchronously at a sample frequency of 100 Hz.

Muscle activity (EMG) was recorded on both limbs of the main leg muscles: m biceps femoris (BF), m. semitendinosus (ST), m. rectus femoris (RF), m. vastus lateralis (VL), m. tibialis anterior (TA), m. gastrocnemius medialis (GM), and m. soleus (SO). Bipolar Ag/AgCl (Medicotest A/S) surface electrodes were attached after cleaning and gentle abrasion of the skin. The center-to-center electrode distance was 2.5 cm. The EMG signals were amplified 20 times (Porti-17™, Twente Medical Systems), high-pass filtered (5 Hz), and stored on disk at a sample frequency of 1000 Hz with a 22-bit resolution. Next, the signals were whitened (fifth order) [14] to reduce the influence of tissue filtering and movement artefacts, Hilbert transformed, rectified and finally low-pass filtered (fifth order Savitzky-Golay filter, frame size of 21).

For each subject, 5 trials of normal walking and 5 trials of forewarned walking were selected with complete EMG and kinematic data. EMG data was normalized with respect to the subject's maximal EMG activity of the averaged normal walking pattern. Heel strike (HS) and toe-off (TO) were detected on the basis of kinematic data [54]. In order to investigate changes in the muscle activity of the walking pattern after forewarning, we first checked whether the variables that were altered in the previous study (i.e. minimum toe height and step width) showed the same differences to assure reproducible effects. Then, we calculated the averaged area under the curve (AUC) of muscle activity for each muscle over a stride. As most trips were on the left side at mid-swing, we also specified the AUC of muscle activity for the left limb during mid-swing over a period of 300 ms (150 ms prior to mid-swing until 150 ms after mid-swing). As the right foot was at mid-stance when most trips occurred, a specified AUC was calculated for the right limb muscles covering 300 ms around mid-stance. All parameters were tested in a multivariate analysis of variance (MANOVA). Comparisons between condition (normal vs. forewarned) involved limb (right or left) and age (young vs. old) and were evaluated univariately for the individual muscles. Significance level was set at p=0.05.

Results

All subjects walked at a constant velocity, but the older subjects walked significantly (p=0.007) slower than the young subjects (Table 3.1). Like in the previous study, both young and older subjects showed an increased foot clearance during mid-swing. In particular, the left foot (the side that was tripped most often), was affected by forewarning in both age groups: in the young subjects, minimum toe elevation of the left foot was increased by 2.3 cm (condition*side: p=0.002), older subjects showed 1.6 cm increase in the minimum toe elevation of the left foot (condition*side: p=0.004). We also found a slight increase (2%) in step frequency and decrease (2%) in stride time in the older subjects (p=0.005 and p=0006, respectively). Velocity tended to increase (2%) in the group of older subjects, but was not significantly affected (p=0.087). Other parameters were not significantly affected by forewarning.

In a general analysis over all young and old subjects of the EMG pattern over a complete stride, a significant difference (p=0.006) was found between normal and forewarned walking. There was no main effect of limb and there were no significant interactions of forewarning with limb or with age. This latter result indicates that the general pattern of changes after forewarning was the same for the young and the older subjects. Univariately, for each muscle separately, the muscle activity pattern was significantly increased after forewarning in RF (12%, p=0.040), VL (11%, p=0.010) and TA (13%, p=0.000). We found a significant interaction of forewarning with age in TA (p=0.014) and with age and limb in RF (p=0.049). These interactions indicate a different pattern between age groups for the TA and the RF muscles. Although the general effects of forewarning on the EMG pattern were the same for the young and older subjects, the interactions for TA and RF suggested that a separate analysis for young and older subjects could provide additional information.

Figure 3.1 shows the averaged EMG patterns during normal walking and forewarned walking for the young subjects. Multivariate analysis of variance showed no significant difference in the group of young subjects over the total stride cycle between the normal and forewarned conditions, nor between limbs. No interaction between condition and limb was found. Univariately, we found significantly increased activity over stride in the forewarned condition in the following muscles: ST (18%, p=0.001), RF (15%, p=0.006), VL (14%, p=0.000) and TA (22%, p=0.001). No side or interaction effects were found. The averaged muscle activity during the specified periods of 300 ms over mid-swing or mid-stance resulted in similar increases after forewarning: ST (16%, p=0.003), RF (21%, p=0.056), VL (17%, p=0.031) and TA (19%, p=0.050). This similar increase over the critical period indicates that the muscle activities were increased equally over the whole gait cycle and that these increases could not be ascribed to an increase in this specified phase.

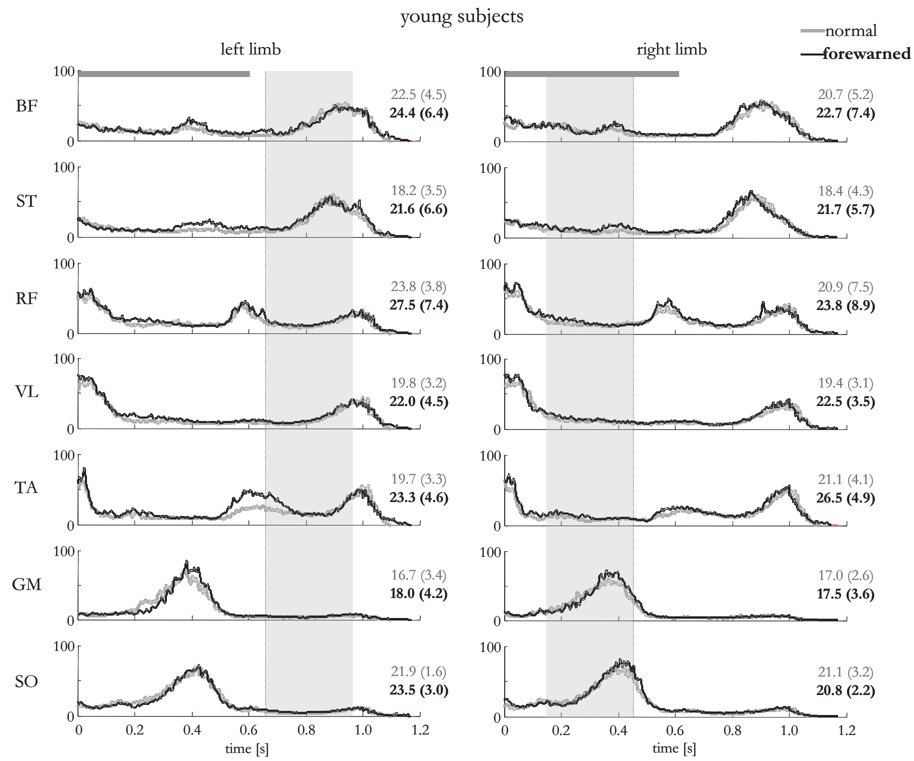


Figure 3.1: *Averaged and normalized muscle activity patterns over one stride (from heel strike to heel strike) for both lower limbs during normal walking (thick gray lines) and forewarned walking (thin black lines) in young subjects. The gray bar at the top represents the stance phase. Shaded areas depict the periods of 300 ms over which the areas under the muscle activity curves were calculated; during mid-swing in the left limb and during mid-stance in the right limb. The averaged areas under the muscle activity curve over these periods (SD) are given for each muscle of each limb for normal walking (in gray) and for forewarned walking (in black).*

Figure 3.2 shows the averaged EMG pattern (over subjects and trials) during normal and forewarned walking for the older subjects. As in the young subjects, the multivariate analysis showed that the overall EMG pattern for the older subjects was not significantly changed after forewarning. No significant differences were found univariately between conditions for the older subjects, except for a significantly increased TA activity in the left limb (17%, p=0.044), the limb that was tripped most often. Specified for the mid-swing and mid-stance phases, no significant differences between condition and interactions were found either. The increases in muscle activity of ST, RF and VL muscles did not exceed 4% of normal AUC. Note the increased variability in area under the EMG curves for the older subjects compared to the young, and that the pattern of changes was the same for young and older subjects (Figures 3.1 and 3.2).

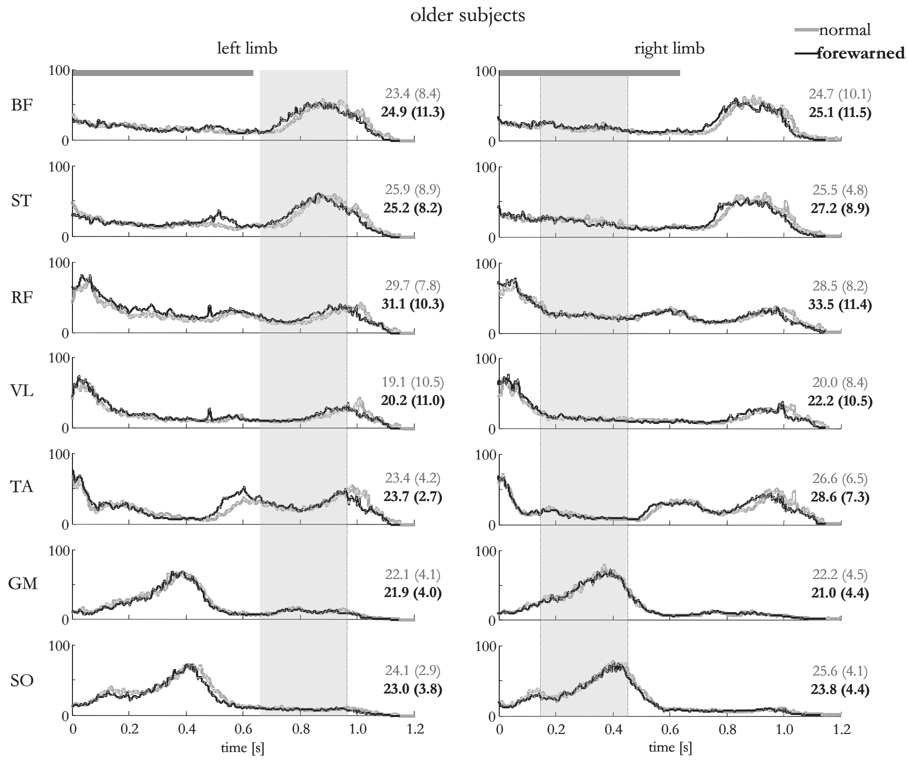


Figure 3.2: *Averaged muscle activity patterns over one stride for both limbs during normal walking (thick gray lines) and forewarned walking (thin black lines) in older subjects. Grey bars and shaded areas are the same as in Figure 3.1. The averaged areas under the muscle activity curve during the shaded periods (and SD) are given for each muscle of each limb for normal walking (in gray) and for forewarned walking (in black).*

Discussion

This study investigated whether anticipatory changes in the muscle activity patterns of lower limb muscles occurred after forewarning of a possible trip in both young and older subjects. No major changes in kinematics were found in the groups of young and older subjects after forewarning, except for an increase of minimum toe clearance during mid-swing. Forewarning did increase the activity in ST, RF, VL and TA muscles for the young subjects, but these increases were small. The older adults showed the same general pattern, but their variability was higher and only the increase in activity of TA of the left limb (which was perturbed most often) was statistically significant. Below we shall discuss the design of the experiment, functional explanations for the observed changes and finally the implications for experiments on perturbations during walking.

Young and older subjects walked over the platform in two conditions. First they were assured that they would not be perturbed, and then they were forewarned for a possible trip, which was actually induced in some trials. In the first walking trial after each trip, we subjectively observed anticipation in most subjects (i.e. avoidance strategy by high lifting of the feet and extreme hip flexion), but after 3 to 5 "catch" trials a relatively normal walking pattern was regained. The selected forewarned trials in this study were trials directly prior to a perturbation trial (on average about 4 trials after a previous trip). These trials are expected to approximate the perturbed pattern and are commonly used in perturbation studies for comparison with perturbed trials.

For comparison of the muscle activity patterns among conditions, we calculated the area under the EMG curve (AUC) over a complete stride cycle. As an averaged stride value did not distinguish the specific time or phase of increased muscle activity, we specified the AUC over the period that was most critical in terms of the possibility of being tripped: in the left limb during mid-swing and in the right limb during mid-stance. Most effect of forewarning was expected during these phases, but statistically, the effects of forewarning during these phases were as high as during the total stride cycle. It could be that the effects were most pronounced in the phases when the muscle is normally active. For example, the increase in TA muscle activity is most pronounced during early swing, when the TA is activated to dorsiflex the foot (Figures 3.1 and 3.2). However, such effects in early swing are not of great importance for responses to trips elicited during mid-swing.

The increased TA activity during swing phase in young and older subjects is in line with the observed increase in minimum toe clearance. This increased activity suggests a strategy to avoid contact with the obstacle, rather than stiffening of the ankle joint, as this would require co-contraction with the triceps surae muscles, which was not observed. The slight increase in ST, RF and VL activity after forewarning could be interpreted as increased co-contraction leading to stiffening of the knee joint. In the older subjects, forewarning did not significantly affect ST, RF and VL muscle activity. It might be that the older subjects did not increase their muscle activity, but it might also be that a small difference between conditions was not significant, due to the increased variance in muscle activity, or due to anxiety in both forewarned and normal walking. This latter effect is unlikely, as it would result in a general interaction between condition and age, and this was not the case. The absence of an interaction between condition and age further points out that forewarning generally had the same effect on the EMG pattern in young as in older subjects.

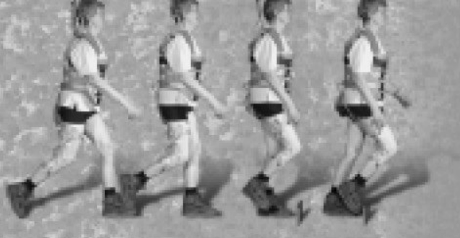
The reason for investigating changes in muscle activity during walking after forewarning was that anticipatory or adaptive changes might occur. Increased activity in antagonistic muscles (i.e. co-contraction) can theoretically result in the same net joint moments, but stiffness and damping can be increased. Consequently, the recovery reactions could be affected (e.g. more pronounced). The only kinematic change we observed was the increased toe elevation at mid-swing (2.3 and 1.6 cm in respectively young and older subjects). This elevation was caused by the increased TA activity and is unlikely to affect the probability of hitting the 15-cm high obstacle in our experiment. As far as tripping responses concerned, it is known from the literature that in the swing limb, early responses (60-80 ms) are seen in the BF, followed by and late responses (110-130 ms) in the RF and VL [20, 74]. In the contralateral stance limb, large and rapid BF, ST and GM bursts (60-80 ms) are observed after tripping [57, 74]. The sizes of these tripping responses are about 100 to 500% of the maximum normal walking activity [20, 57, 74]. The TA muscle shows variable onsets and activities (facilitated or suppressed responses) [20, 74], and of much smaller amplitudes. In the present study, we found a clear tendency towards increased activity in antagonistic muscles after forewarning. This tendency was strong enough to result in significant effects in some muscles (i.e. ST, RF, VL and TA), but generally, these increases were only about 11% (ranging from 4 to 22%) of the normal walking activity. These baseline activity increases are not expected to have substantial effects on the magnitude of the large perturbation responses in young or older subjects.

Conclusion

After forewarning of a possible trip, statistically significant increases in muscle activity can be discerned in the walking patterns of young and older subjects. These effects, however, are only marginal when compared to the magnitude of tripping responses. Therefore, valid experimentation with respect to tripping reactions is possible, provided a number of "catch" trials.

Contribution of the support limb in control of angular momentum after tripping

Journal of Biomechanics
in press



4

Chapter

Abstract

Tripping over an obstacle can result in a fall when the forward angular momentum, obtained from impact with the obstacle, is not arrested. Angular momentum can be restrained by proper placement of the recovery limb, anteriorly of the body, but possibly also by a reaction in the contralateral support limb during push-off. The purpose of this study was to quantify the extent to which the support limb contributes to recovery after tripping by providing time and clearance for proper positioning of the recovery limb, and by restraining the angular momentum of the body during push-off. Twelve young adults were repeatedly tripped over an obstacle during mid-swing, while walking over a platform. Kinematics and ground reaction forces at the support limb were measured. Quantification of angular momentum was based on calculation of the external moment, which equals the rate of change in the angular momentum of the body. Results showed that all subjects acquired a similar increase in angular momentum during foot-obstacle contact, on average 11.4 $kg \cdot m^2 \cdot s^{-1}$. In all subjects, the support limb played a role in recovery after tripping by providing time and clearance for proper positioning of the recovery limb, as indicated by body elevation (6%) and the increased forward pelvis displacement over recovery stride (43%). Almost all subjects were also able to restrain the forward angular momentum of the body during push-off by the support limb. Less angular momentum remained to be further accomplished by the recovery limb. Reductions in the quality of the support limb responses may be among the factors that increase the risk of falling in the elderly.

Introduction

Falls and fall-related injuries cause serious problems for the growing population of the elderly. One in three adults over 65 years of age falls once a year, mostly as the result of a trip or slip [2, 44, 70]. The need to discover mechanisms underlying trip-related falls has led to several investigations of tripping [20, 26, 27, 51, 73, 74, 78].

The main purpose of the recovery reaction after tripping is to arrest the forward angular momentum, which the body gets from impact with the obstacle. An inadequate reaction will lead to a fall. Eng et al. [20] described two phase-dependent modes of recovery reactions. Impact during early swing leads to an elevating strategy, in which the obstructed (ipsilateral) swing limb is lifted over the obstacle immediately after collision and placed forward, over the obstacle. Impact during late swing induces a lowering strategy, in which the obstructed foot is placed quickly before the obstacle and the other limb is subsequently placed anteriorly of the body. For both strategies, we call the limb that is placed anteriorly of the body the recovery limb, while the contralateral stance limb is called the support limb.

Placing the recovery limb anteriorly of the body is one means to reduce the angular momentum of the body [26, 27, 51]. This limb can generate a force and moment that counteract the angular momentum, provided that it is properly placed anteriorly of the body. Proper placement of the recovery ling can only be achieved if there is sufficient time and clearance. This can be brought about by rapid responses in the recovery limb itself, but in addition, the support limb can help to gain time and clearance by elevating the body during push-off.

In theory, the support limb can also contribute to recovery in another way, namely by reducing the forward angular momentum of the body during push-off, before the recovery limb hits the ground. Angular momentum can be controlled by generating adequate joint moments, and the associated rate of change in angular momentum is reflected in the external moment (M_{ext}), which is the moment of external forces about the body center of mass.

The purpose of this study was to determine whether the support limb contributes to recovery after tripping, and if so, to quantify the extent to which it contributes. We hypothesized the support limb to contribute in two ways: (a) by providing time and clearance for proper positioning of the recovery limb and (b) by restraining or reducing the forward angular momentum of the body induced by the trip. The first role would be reflected in an increased upward and forward displacement of the pelvis during the push-off phase in tripping as compared to normal walking. The second role would be reflected in a sign change in the external moment during the push-off phase.

Methods

Twelve volunteers (6 male, 6 female) with a mean age of 27 years (SD 4) participated in this study. Subjects were informed on the research procedures before they gave informed consent in accordance with the ethical standards of the declaration of Helsinki. Participants walked approximately 60 times over a platform in which 21 obstacles were hidden. In about 10 trials, the subjects were tripped over one of these obstacles. A computer controlled, based on online kinematic data, which one of these obstacles had to appear at what time, so as to cause a trip at mid-swing, allowing us to focus on the elevating strategy. In addition to kinematics, we measured ground reaction forces of the support limb. Details on the experimental setup and protocol are described below.

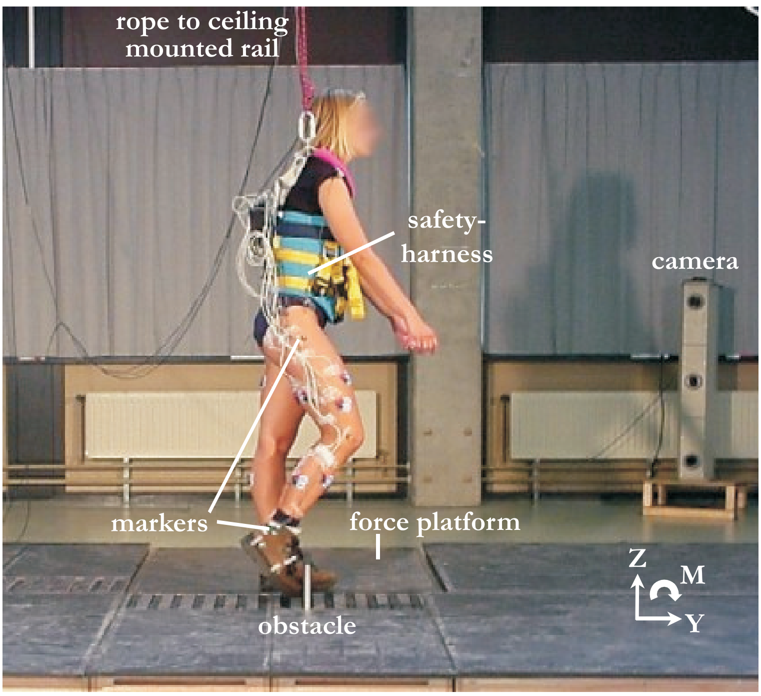


Figure 4.1: *Picture of the experimental setup. A force platform and Optotrak cameras were used for data collection of kinetics and kinematics. Twenty-one obstacles were hidden in the floor. One obstacle could suddenly appear, based on kinematic data of the ongoing trial, to trip the subjects at a specific time. Subjects wore a safety-harness.*

Subjects, wearing walking shoes, were instructed to walk at a self-selected speed over a platform of 12 meters. In the platform, a force plate was mounted and 21 aluminum obstacles of 15-cm height (28.5-cm width) were hidden over a total distance of 1.5 m (Figure 4.1). In about 10 out of 60 walking trials, one of the obstacles suddenly appeared to trip the subject, either on the left or the right side. At the start of each trial, subjects did not know whether, or where an obstacle would appear. Online kinematic data of each trial were used to calculate the subject's step length and velocity. Based on these variables, position and timing of the obstacle to appear were chosen, so as to cause a trip at a certain percentage of the swing phase. Given the inter-obstacle distance of 7 cm, the obstacle appeared within 3.5 cm of the calculated position. The experimenter controlled whether or not an obstacle should appear, at which side (left or right) and at which percentage of the swing phase. In this experiment, at least 5 trips were evoked to trip the subject on the left limb at mid-swing to obtain comparable reactions (elevating strategy) and collect the ground reaction force data while the support limb was on the force platform. A full-body safety harness, attached to a ceiling-mounted rail, ensured that subjects would not be injured should their recovery reaction be inadequate. The safety ropes provided enough slack for free motion and harness assistance could be precluded visually, to which end all trials were recorded on video.

Gait kinematics were recorded during each trial using 4 Optotrak cameras (Northern Digital ©). Motion of 12 infrared-light emitting markers was tracked. The markers were placed bilaterally over the anatomical landmarks heel, metatarsophalangeal joint (MTP5), lateral malleolus, lateral epicondyle and trochanter major of the femur, and acromial process. The coordinates of these landmarks defined 7 body segments: 2 feet, 2 lower legs, 2 upper legs and a head-arms-trunk (HAT) segment. Ground reaction forces at the right foot were recorded by a custom-made strain gauge force plate (1x1m). From the distribution of the force components, the center of pressure (COP) was calculated. LabVIEW (National Instruments ©) was used to synchronize and collect the kinematic data and ground reaction forces at a sample frequency of 100 Hz and to control the appearance of obstacles hidden in the walkway (see above and [45]).

For each subject, 5 normal walking trials and 5 left leg tripping trials at mid-swing were randomly selected from successful trials with complete kinematic and dynamic data. In 2 subjects, complete data of only 3 tripping trials was available.

Heel strike (HS) and toe-off (TO) were detected on the basis of kinematic data, as force plate data were not available for the left foot. HS coincided with a local minimum in the vertical velocity component of the toe marker and TO coincided with a local maximum in the vertical velocity component of the heel marker [54]. Impact (or contact) of the foot with the obstacle coincided with a local minimum in the acceleration of the toe marker in the walking direction. Based on HS, TO and obstacle-foot contact events, data were analyzed in the sagittal plane after smoothing with a one-directional second order low-pass Butterworth filter with a cutoff frequency of 8 Hz. One-directional filtering preserved the timing of the start of obstacle-foot contact onto the data.

To investigate the contribution of push-off by the support limb in gaining time and clearance for proper positioning of the recovery limb, we calculated body elevation (hip height) and timing parameters. Hip height was calculated as the height of the bilateral hip markers, relative to subjects' hip height at HS. For timing parameters, we calculated duration of stride (from HS until HS), stance phase (from HS until TO), swing phase (from TO until HS) and double support phase (from HS of the one limb until TO of the other limb). For statistical analysis of differences in these parameters between normal walking and tripping reactions, within-subject averaged (across trials) values were analyzed in a multivariate analysis of variance (MANOVA) for repeated measures. The level of significance was set at p=0.05.

$$M_{ext} = F_{gr} \times d_{gr} + F_c \times d_c$$

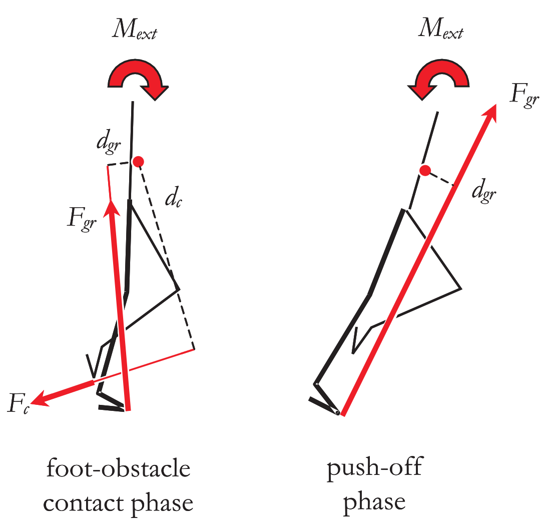


Figure 4.2: *External moment (M_{ext}) was calculated as the sum of moments of the external forces about the body center of mass (●). F_{gr} is the ground reaction force and F_c is the foot-obstacle contact force, d_{gr} and d_c (dashed lines) are the moment arms of the respective force vectors. During foot-obstacle contact phase, the moment effect of F_{gr} and F_c on the body indicates an increase in forward angular momentum, whereas during push-off, theoretically, a decrease in angular momentum can be achieved.*

The contribution of the support limb to restrain angular momentum of the body during push-off was investigated by calculating the external moment (M_{ext}), which equals the rate of change in the angular momentum of the entire system. Calculation of angular momentum directly from the kinematic data was not deemed to be very accurate, because the angular momentum of arm segments, which made vigorous flexion and endorotation, could not be determined. Therefore, M_{ext} was used as a measure for the rate of change in angular momentum.

M_{ext} was calculated as the sum of the moments generated by external forces acting on the system:

$$M_{ext} = \frac{d\sum(I\omega)}{dt} = \vec{F}_{gr} \times \vec{d}_{gr} + \vec{F}_c \times \vec{d}_c \tag{4.1}$$

where $\vec{F}_{gr}$ is the ground reaction force at the COP, $\vec{F}_c$ the contact force of the obstacle at the toe, $\vec{d}_{gr}$ and $\vec{d}_c$ are the vectors from the body center of mass (COM) to the point of application of the respective force vectors (Figure 4.2). GRF was measured directly by the force platform. The obstacle-foot contact force (F_c) was calculated from the linear impulse over a period from 10 ms prior to impact to return of the foot-obstacle contact force to 0 N. During this phase, the external contact force equals the difference between the rate of change in linear impulse and the GRF plus force of gravity:

$$\vec{F}_c = \frac{d(m_{body} \cdot \dot{\vec{R}}_{COM})}{dt} - \vec{F}_{gr} - m_{body} \cdot \vec{g} \tag{4.2}$$

where $\vec{F}_c$ and $\vec{F}_{gr}$ are the contact forces and GRF, respectively, m_{body} is body mass, $\dot{\vec{R}}_{COM}$ is the linear velocity of the body COM and $\vec{g}$ is -9.81. The contact phase was followed by the push-off phase, which is defined as the period from the end of foot-obstacle contact to the end of the single support phase.

For determination of the vectors $\vec{d}_{gr}$ and $\vec{d}_c$, the position of the body COM was calculated from the segments' masses and center of mass locations. The inertial parameters of each segment (mass, position of the segmental center of mass and the segmental moment of inertia) were calculated per subject, according to Plagenhoef [61]. The HAT was represented as a single link from the bilateral hip joint centers to the HAT COM. The position of the HAT COM was calculated by using the criterion that the reactive forces acting at the hips, calculated by inverse dynamics, equaled the force necessary for (translational) acceleration of the HAT segment:

$$\ddot{\vec{R}}_{COM} = \frac{\vec{F}_{hips} + m_{body} \cdot \vec{g}}{m_{body}} \tag{4.3}$$

where $\vec{F}_{hips}$ is the force acting at the hips. This way, we calculated the acceleration of the HAT COM, resulting in zero residual forces. Velocity and position of HAT COM were calculated by integration. For initial conditions we used velocity and position of a HAT COM on the line between hip and shoulder joint centers at the first sample of the single support phase. Calculation of HAT COM was limited to the single support and aerial phases, as the external GRF was only available for this period.

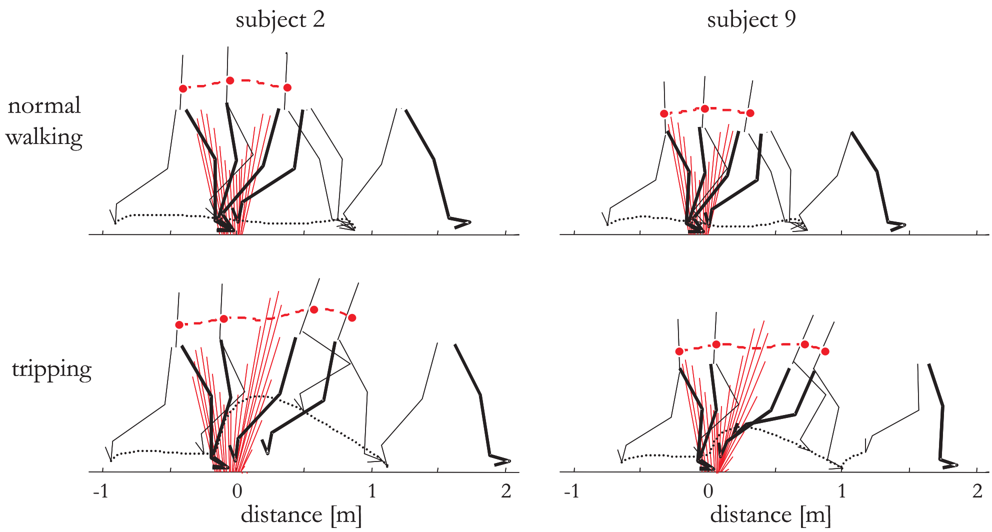


Figure 4.3: *Stick figures of two subjects (left or right columns) during a typical walking trial (upper graph) and a tripping trial (lower graph) for 5 instants of time (right and lift toe-off and heel strike, mid-swing or trip initiation). The obstructed (left) swing limb is indicated by thin lines, thick lines depict the contralateral support limb. The HAT segment is defined from the bilateral hip joint centers to the (optimized) location of HAT COM. Ground Reaction Force vectors, body COM position (●) and trajectory (dashed line), as well as toe trajectory over time (dotted line) are drawn. Note that HAT COM could not be calculated during double support.*

Results

Tripping reactions were induced on average at 39 (SD 3.8) % of the normal swing phase duration. Typically after tripping in this particular phase of the gait cycle, subjects performed an elevating strategy. Figure 4.3 depicts stick diagrams of two typical subjects (2 and 9) for both normal walking and tripping.

Table 4.1: *General parameters (timing, stride length and bilateral hip height) during normal walking (averaged over limbs) and tripping (for support limb and obstructed swing limb separately). Averages (and SD) over 5 trials and 12 subjects. Negative double support indicates an aerial phase.*

	normal walking: left & right limbs	tripping: support limb (push-off)	tripping: swing limb (recovery)	
velocity (m/s)	1.61 (0.15)	1.61 (0.17)	1.44 (0.14)	* # §
frequency (steps/min)	117 (4.50)	109 (12.27)	96 (7.71)	* # §
cycle time (s)	1.03 (0.04)	1.12 (0.12)	1.26 (0.10)	* # §
stance phase (s)	0.61 (0.03)	0.69 (0.07)	0.61 (0.03)	* # §
swing phase (s)	0.44 (0.02)	0.45 (0.08)	0.67 (0.09)	* # §
double support (s)	0.09 (0.01)	0.09 (0.01)	-0.05 (0.05)	* # §
stride length (m)	1.66 (0.15)	1.81 (0.26)	1.83 (0.24)	*
hip height at toe-off (m)	0.86 (0.04)	0.91 (0.04)		*
hip displacement over stride (m)	0.83 (0.08)	1.19 (0.16)		*

** significant difference between conditions; # difference between sides; § interaction condition × side*

Table 4.1 represents the general parameters for both normal walking and tripping. The duration of a stride, normally 1.03 (SD 0.04) s, was increased significantly for the obstructed swing (recovery) limb as well as for the support (push-off) limb. The increase in stride duration was attributed to an increase in stance phase duration of the push-off limb (13%), and to an increase in swing phase duration of the recovery limb (63%). The double support phase was not present after tripping. Instead, an aerial phase was seen. These findings indicate that extra time was available for positioning of the recovery limb. Furthermore, the stride length of the recovery limb was increased (10%). Stride length can be determined by actions of both the support limb and the recovery limb, but horizontal displacement of the pelvis over the recovery stride (i.e., from toe-off until landing of the recovery foot) can only be achieved by actions in the support limb. The bilateral hip displacement was increased by about 43% after tripping (Table 4.1).

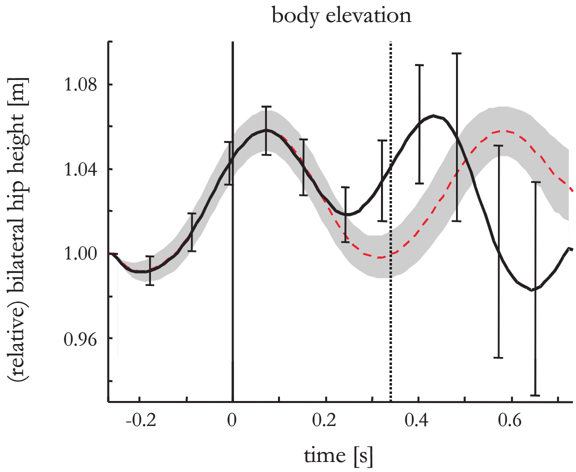


Figure 4.4: *The relative height of the bilateral hip markers as indication for body elevation during push-off, averaged over trials and subjects, and relative to subjects' hip height. Mean graphs over complete stride (from heel strike to heel strike) for normal walking (dashed mean, shaded SD) and for tripping (solid line, SD in error bars). Vertical lines indicate trip initiation (solid) and end of single support phase (dotted).*

Furthermore, the body was elevated during push-off, as can be seen in Figure 4.4. During normal walking, the position of the bilateral hip joint markers is highest in mid-stance and lowest in the double support phase, whereas after tripping, the body was elevated additionally during push-off by the support limb. This was seen in all subjects. At the end of push-off, the averaged hip height was about 5 cm higher after tripping compared to normal walking (Table 4.1). Typical obstacle-foot contact forces are presented in Figure 4.5. Contact duration, averaged over trials and subjects, was 115 (SD 20) ms (Table 4.2). The horizontal (fore-aft) peak force was on average –177 (SD 43) N. The vertical force showed in all subjects a maximum of on average 48 (SD 24) N, followed by a minimum of –84 (SD 45) N.

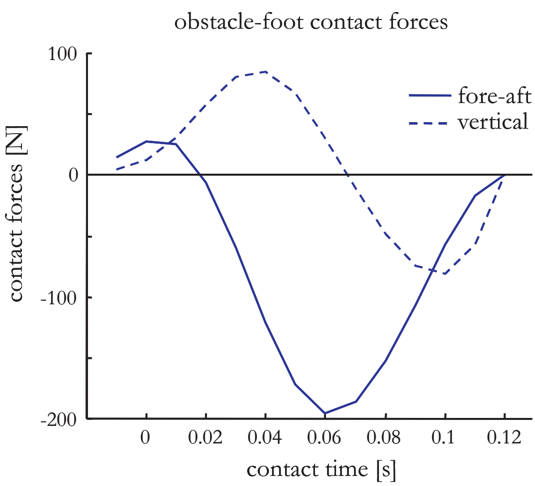


Figure 4.5: *Obstacle-foot contact forces (fore-aft and vertical) from 10 ms prior to impact to return of contact forces back to 0 N, for one typical tripping trial.*

During normal walking, a propelling M_{ext} is generated during push-off and an upright position is maintained by a counteracting force at heel strike of the next step. Successive positive and negative excursions in M_{ext} cancel each other over a stride cycle. Until trip initiation, there was of course no difference in M_{ext} between the walking and tripping conditions. During obstacle-foot contact, the body started rotating forward (clockwise), due to the external contact forces and gravity. The increase in angular momentum is reflected in a positive M_{ext} (i.e., angular acceleration of forward rotation). In all subjects, the area under the curve (AUC) of M_{ext}, which equals the angular momentum, was increased over obstacle-foot contact phase, by on average 11.4 kg $\cdot m^2 \cdot s^{-1}$ (Table 4.2).

Table 4.2: *Duration (ms) of obstacle-foot contact (from collision to obstacle free), push-off (from obstacle-free to end single support phase) and the sum of both phases. Averages (and SD) per subject over 5 trials. Area under the curve (AUC, kg $\cdot m^2 \cdot s^{-1}$) of M_{ext} over obstacle-foot contact phase, push-off phase and the sum of both phases. Subject 1 to 4 were able to fully reduce the increased angular momentum (negative, counterclockwise AUC of M_{ext} during push-off), subject 5 to 10 restrained the increase (AUC of about 0 kg $\cdot m^2 \cdot s^{-1}$), and subject 11 and 12 were not able to restrain during push-off (further positive AUC of M_{ext} over push-off).*

subject	contact duration	push-off duration	total duration	contact AUC M_{ext}	push-off AUC M_{ext}	total AUC M_{ext}
1	96 (23)	342 (22)	438 (12)	11.2 (2.4)	-14.5 (1.4)	-3.3 (2.2)
2	120 (26)	254 (20)	374 (21)	14.1 (3.8)	-12.8 (3.8)	1.3 (1.6)
3	108 (15)	194 (44)	302 (36)	14.0 (2.4)	-7.8 (3.5)	6.2 (5.7)
4	120 (13)	228 (41)	348 (33)	8.5 (1.6)	-6.3 (4.6)	2.2 (3.1)
5	110 (15)	262 (30)	372 (15)	12.1 (1.8)	-1.9 (4.3)	10.2 (5.8)
6	114 (5)	358 (44)	472 (42)	9.4 (2.3)	-1.6 (1.7)	7.8 (3.3)
7	110 (0)	327 (29)	437 (29)	10.0 (0.5)	0.6 (3.6)	10.6 (3.1)
8	132 (7)	322 (23)	454 (26)	14.5 (1.8)	0.6 (2.3)	15.1 (3.1)
9	132 (16)	274 (24)	406 (19)	11.4 (3.2)	1.1 (1.3)	12.5 (3.2)
10	132 (4)	394 (47)	526 (48)	17.1 (1.4)	6.1 (2.9)	23.3 (3.0)
11	93 (17)	193 (29)	287 (38)	6.0 (1.7)	6.3 (2.3)	12.3 (3.4)
12	100 (14)	278 (32)	378 (45)	8.5 (1.3)	11.0 (0.8)	19.5 (1.1)
all	115 (20)	287 (69)	402 (73)	11.4 (1.6)	-2.0 (2.7)	9.8 (3.2)

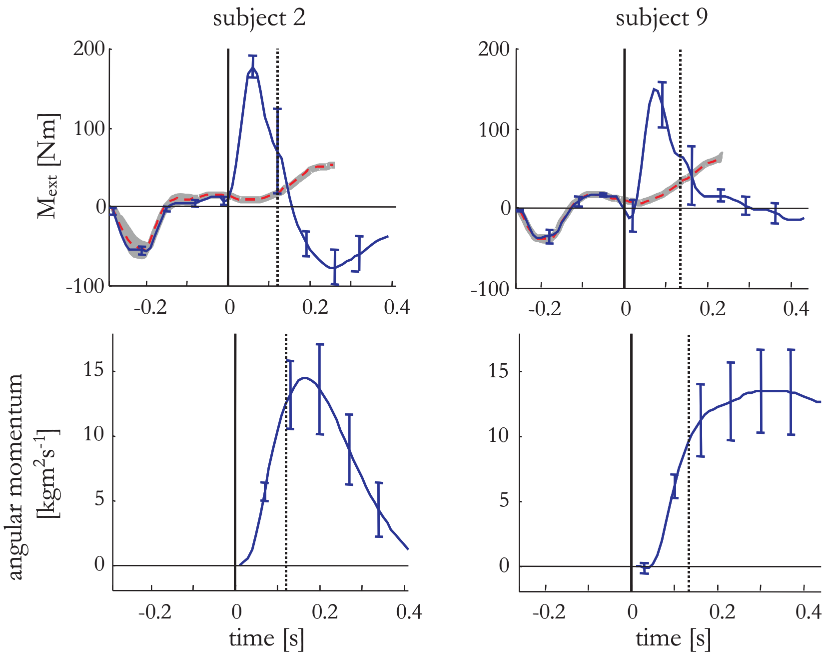


Figure 4.6: *M_{ext} and integral of M_{ext} which equals the angular momentum. Graphs from heel strike of the support limb until end of single support phase and averaged over 5 trials for normal walking (dashed mean and shaded SD) and for tripping (with error bars). Vertical lines indicate trip initiation (solid) and end of foot-obstacle contact phase (dashed). A positive M_{ext} reflects an increase of angular momentum (clockwise acceleration), a negative M_{ext} indicates a decrease of angular momentum (counterclockwise).*

Angular momentum can be controlled by generating adequate joint moments during push-off; the decrease in angular momentum would be reflected in a negative external moment. When considering the contribution of the support limb to recovery of the angular momentum, substantial between-subject variations were noted, although the reproducibility within-subjects seemed very high (see SD in Figure 4.6 and Table 4.2). Three subgroups could be defined, based on the capacity to restrain the angular momentum during push-off (Table 4.2). For the two major subgroups, individual data of two representative subjects will be presented first and described in detail. Figure 4.6 presents the external moments and the integral of M_{ext}, averaged over trials of these two exemplary subjects. M_{ext} in subject 2 became negative during push-off, indicating that the angular momentum is reduced. In this subject, the forward rotation that the body acquired during impact is completely eliminated during push-off (Table 4.2). Subject 9 stopped the increase in angular momentum as well, but did not manage to reduce it. This subject needed an extra step of the recovery limb to fully eliminate the forward rotation of the body. Indeed, Figure 4.3 shows another jump and aerial phase in the following step for subject 9, whereas subject 2 had regained a normal walking pattern in the subsequent step. The outcomes of the reactions by the two subjects presented were representative for the main subgroups (Table 4.2 and Figure 4.7). Elimination of the angular momentum was achieved by 4 subjects (numbers 1 to 4) and reduction was achieved by 6 subjects (numbers 5 to 10). Two subjects (11 and 12) did not react adequately during push-off; their angular momentum continued to increase over the whole push-off duration. Still, none of the subjects fell into the harness, so they were all able to recover eventually, although contribution of support limb was different and subsequent recovery steps were necessary in some subjects.

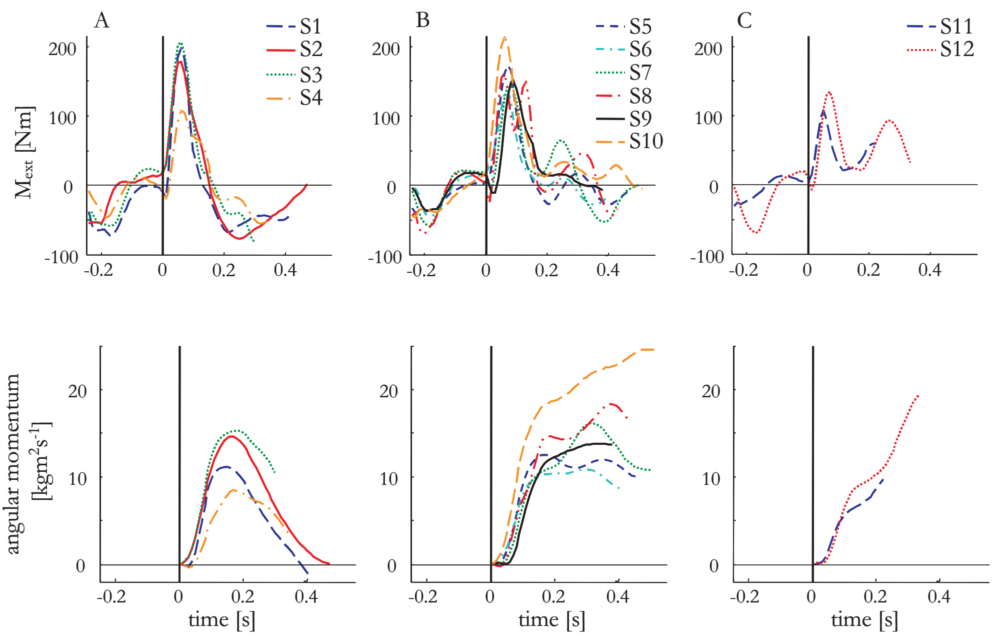


Figure 4.7 *M_{ext} and angular momentum for all subjects, divided in the 3 subgroups (see Table 4.2): A) group of 4 subjects who were able to fully reduce the angular momentum (angular momentum back to zero over push-off), B) group of 6 subjects who restrained the increase in angular momentum (integral becomes constant), and C) 2 subjects who were not able to restrain during push-off (further increase in angular momentum over push-off).*

Discussion

This study revealed the contributions of the support limb to recovery after tripping. In all subjects push-off generated by the support limb provided extra time and clearance for proper positioning of the recovery limb. In most subjects the support limb additionally contributed by restraining or reducing the angular momentum of the body during the push-off. Up to date the literature on tripping has mainly focused on the swing limb [20, 26, 51, 73, 74]. The present results suggest that support limb responses are functionally important and merit further investigation. Before discussing the role of the support limb in recovery after tripping, we need to address some methodological points.

The results presented here were based on experiments in which subjects were aware of the fact that they would be tripped in some trials. We have previously shown that this does not greatly affect gait kinematics [54]. The high reproducibility of the characteristics of the tripping responses in the present study (see Figure 4.6 and Table 4.2) supports the idea that valid experimentation with respect to tripping responses is possible.

Estimation of the external moment (M_{ext}), which was used to study the angular momentum, required knowledge the location of the body COM. As in earlier studies [32], we used optimization methods to improve the position of the trunk COM to get a better body COM. Acceleration of the HAT COM was based on the reactive forces acting at the hips, which reflect (translational) accelerations of the HAT segment. In the first phase of single support, we assume no effect of arm swing on HAT COM and therefore we felt safe to use velocity and position of a HAT COM on a fixed point on the line between hip and shoulder joint centers (according to Plagenhoef [61]) for initial conditions for integration.

Another requirement for the validity of calculation of M_{ext} was the determination of the obstacle-foot contact force. We based the calculation of these contact forces on the linear impulse. Over the duration of obstacle-foot contact, we expected no effects yet from arm movements on linear velocity of body COM. Recently, Zhou et al. [109] measured obstacle-foot contact forces during walking, using a 3D-force platform. They found a contact duration of 90 ms, with a fore-aft and vertical maximum value of 129 and 49 N, respectively. Our calculations yielded similar contact duration and peak values of forces. Any difference might be caused by a difference in walking velocity and time of impact during the swing phase of single trial measurements of Zhou et al. [109].

Quantification of the angular momentum by calculation of M_{ext} enabled us to investigate the contribution of the support limb to recovery after tripping. All subjects showed a similar increase in angular momentum during foot-obstacle contact. Provided proper (forward) positioning of the recovery limb, this limb can generate a force and moment that counteract the angular momentum of the body. After being tripped, all subjects showed an increase in stance duration of the support limb and swing duration of the recovery limb, an aerial phase instead of double support, as well as body elevation during push-off and elongation of the stride. Eng et al. [20] also mentioned body height elevation during the elevating strategy. Body elevation started early in the push-off phase (Figure 4.4). Rapid body elevation and forward propelling of the pelvis, together with the duration of stance, swing and aerial phase indicated that push-off by the support limb contributed to gaining time and clearance for proper placement of the recovery limb.

During push-off adequate joint moments in the support limb (reflected in M_{ext}) can also contribute to a reduction of angular momentum. Although results were very reproducible within subjects, different reactions were noted among subjects. Almost all subjects restrained the angular momentum after tripping by a reaction of the support limb, but not all subjects were able to actually reduce the angular momentum during this phase. The question remains what caused these differences between subjects. It could be due to initial conditions, such as walking velocity, trunk angle (and velocity) at time of tripping or joint moment generating capacity. However, no such differences between subjects in walking velocity or trunk angle were obvious. It seems, therefore, that the quality of the reaction in the support limb differs among subjects. Still, the present study showed that all subjects reacted very rapidly in an attempt to control the angular momentum. Further research on response times and response mechanics is required to investigate how an adequate push-off reaction is achieved.

The results of the present study show that the support limb plays an important role in recovery after tripping during push-off. For proper placement, the obstructed swing limb, of course, has to be swung forward. Mechanical requirements in the recovery limb, however, are first expected to become critical after landing when forces and moments have to be generated for counteraction of the angular momentum of the body. The support limb can provide enough time and clearance for proper positioning of the recovery limb. Furthermore, the more reduction in angular momentum achieved by the support limb during push-off, the less remains to be accomplished by the recovery limb. All subjects provided time and clearance during push-off. Most subjects were also able to restrain angular momentum of the body during the push-off by the support limb; some of them even completely reduced the forward angular momentum. Reductions in the quality of the support limb responses may be among the factors that increase the risk of falling in the elderly. Further research is needed to characterize these responses in both young and elderly subjects.

Acknowledgements

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How early reactions in the support limb contribute to balance recovery after tripping

Chapter 5

Journal of Biomechanics
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Abstract

Tripping causes a forward angular momentum that has to be arrested to prevent a fall. The support limb, contralateral to the obstructed swing limb, can contribute to an adequate recovery by providing time and clearance for proper positioning of the recovery limb, and by restraining the angular momentum of the body during push-off. The present study investigated *how* such a contribution is achieved by the support limb in terms of response times and muscle moment generation, in order to provide more insight into the requirements for successful recovery after tripping. Twelve young adults repeatedly walked over a platform in which 21 obstacles were hidden. Each subject was tripped over one of these obstacles during mid-swing in at least 5 trials. Kinematics, dynamics and muscle activity were measured. Very rapid responses were seen in the muscles of the support limb (~65 ms), causing fast increases in muscle moments in the joints during the primary phase of recovery. Especially a large ankle plantar flexion moment (204 Nm), a knee flexion moment (-54 Nm) and a hip extension moment (52 Nm), generated by triceps surae and hamstring muscle activity, brought about the necessary push-off reaction and simultaneously caused a restraining of the forward angular momentum of the body. These required joint moments could be a problem for the elderly who might not be able to rapidly generate such powerful moments. Strength training in these muscle groups may be indicated in elderly subjects to reduce the risk of falling after a trip.

Introduction

Tripping is one of the main causes for falls and fall-related injuries, especially in the population of elderly [2, 44]. Investigation of the mechanisms underlying recovery after tripping can provide insight into balance control. Moreover, understanding how successful recovery is achieved might help to identify causes for inadequate reactions and falls, particularly in elderly people with a high risk of falling. This has motivated several investigations into the organization of recovery reactions following a trip [20, 26, 27, 47, 50-52, 56, 72, 74, 78].

Eng et al. [20] described two phase-dependent modes of recovery reactions: an elevating and a lowering strategy. In the elevating strategy the obstructed (ipsilateral) swing limb is elevated immediately after collision to continue the ongoing step. In the lowering strategy the obstructed foot is placed before the obstacle and the other limb is subsequently placed behind the obstacle. Recovery foot is in both strategies the foot that is positioned behind the obstacle. The contralateral stance limb is called the support limb.

The primary phase of the tripping response, from impact with the obstacle until placement of the recovery foot, was coined positioning phase by Grabiner et al. [26]. The term positioning phase suggests that the essence of this phase is to position the recovery foot by reactions in the recovery limb. Indeed, placing the recovery foot properly anteriorly of the body center of mass is one means to reduce the angular momentum, which the body gets from impact with the obstacle. When properly placed, the recovery limb can generate a force and moment that counteract the body angular momentum [27]. However, other actions can also contribute substantially to an adequate recovery during this primary phase. First of all, the obstructed foot has to be released from the obstacle, about which the body rotates during contact. Furthermore, while the recovery limb is positioned, but before it hits the ground, a strong push-off reaction is seen in the support limb [56]. This response can play a major role in recovery after tripping. By pushing-off with the support limb, time and clearance is provided for proper positioning of the recovery limb. Moreover, generation of adequate joint moments in the support limb restrains the angular momentum of the body before the recovery limb hits the ground.

Although the importance of the role of the support limb in recovery after tripping has been described [56], the characteristics of the primary reactions in this limb are still unknown. The present study was designed to investigate how the support limb contributes to a successful recovery. Important factors for success are response time and the quality of the executed response [27]. Thus far, response quality was described in terms of activation patterns and kinematics, mainly of the recovery limb [20, 26, 27, 51, 73, 74]. The present study will focus on muscle responses and moment generation in the support limb, as these could be limiting factors for recovery in the elderly.

The purpose of this study was to determine how the support limb contributes during push-off to recovery after tripping, in order to provide more insight into the requirements for successful recovery. For this purpose, we had 12 young subjects walk over a platform, and tripped them several times. Kinematics, ground reaction forces and muscle activity were measured and the response times and response mechanics during push-off by the support limb were investigated. We hypothesized that the support limb would show rapid responses, high peak muscle moments and high rates of change in moments generated during push-off. If rapid and strong reactions are required, these could be limiting factors for recovery in the elderly.

Methods

Twelve volunteers (6 male, 6 female) with a mean age of 27 years (SD 4) participated in this study. Subjects were informed on the research procedures before they gave informed consent in accordance with the ethical standards of the declaration of Helsinki. Subjects, protocol and data collection were identical to those described in Pijnappels et al. [56]. Subjects, wearing walking shoes, were instructed to walk at a self-selected speed over a 12 by 2.5 m platform. In the platform, a force plate was mounted and 21 aluminum obstacles (15 cm height) were hidden over a total distance of 1.5 m (Figure 4.1). In about 10 of 60 walking trials, one of the obstacles suddenly appeared to catch the swing leg of the subject. At the start of each trial, subjects did not know whether or where an obstacle would appear. Online kinematic data was used to calculate where and when an obstacle had to appear to initiate a trip at mid-swing. A full-body safety harness, attached to a ceiling-mounted rail, prevented subjects from falling on the floor. The safety ropes provided enough slack for free motion and harness assistance could be precluded visually, to which end all trials were recorded on video.

Gait kinematics were recorded using 4 Optotrak cameras (Northern Digital ©). Motion of 12 infrared-light emitting markers was tracked. The markers were placed bilaterally on the anatomical landmarks heel, 5th metatarso-phalangeal joint, lateral malleolus, lateral epicondyle and major trochanter of the femur, and acromial process. The coordinates of these landmarks defined 7 body segments: 2 feet, 2 lower legs, 2 upper legs and a head-arms-trunk segment. Ground reactions forces (GRF) and center of pressure of the support limb were measured with a custom-made strain gauge force plate (1x1 m). Kinematic and ground reaction force data were collected and synchronized at a sample frequency of 100 Hz. Data were analyzed in the sagittal plane after smoothing with a uni-directional second order low-pass Butterworth filter with a cutoff frequency of 8 Hz. Joint forces and moments were calculated using an inverse dynamics model. For each segment, mass, center of mass position and the moment of inertia were calculated per subject, according to Plagenhoef [61].

For measurement of muscle activity patterns, bipolar Ag/AgCl (Medicotest A/S) surface electrodes were attached after cleaning and gentle abrasion of the skin. The center-to-center electrode distance was 2.5 cm. The electromyogram (EMG) signals were recorded from m. gluteus maximus (GL), m biceps femoris (BF), m. semitendinosus (ST), m. rectus femoris (RF), m. vastus lateralis (VL), m. gastrocnemius medialis (GM), m. soleus (SO), and m. tibialis anterior (TA). The EMG signals were amplified, high-pass filtered (5 Hz), sampled at 1000 Hz and stored on disk (Porti-17™, Twente Medical Systems). Next, the signals were whitened (fifth order) [14] to reduce the influence of tissue filtering and movement artefacts, Hilbert transformed and rectified and finally low-pass filtered (fifth order Savitzky-Golay filter, frame size of 21). This filtering method preserves sudden activity onset without producing a phase-lag. Figure 5.1 illustrates the effects of these techniques.

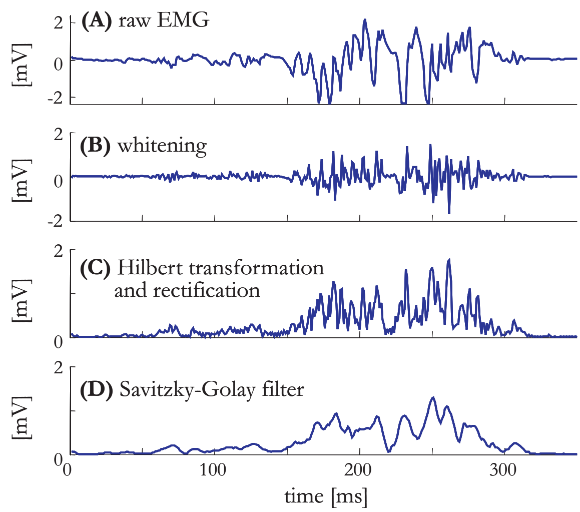

Figure 5.1: *EMG data processing: (A) raw EMG signal of BF muscle activity, (B) the signal after the whitening procedure, (C) Hilbert transformed and rectified, and (D) low-pass filtered by use of a Savitzky-Golay filter.*

For each subject, 5 normal walking and 5 left-leg tripping trials at mid-swing were randomly selected from trials with complete kinematic and dynamic data. In 2 subjects only complete data of 3 tripping trials were available. Heel strike (HS) and toe-off (TO) were detected on the basis of kinematic data [54]. Impact of the foot with the obstacle coincided with a local peak in first derivative of acceleration, also known as jerk, of the toe marker in the walking direction. This kinematic method of detecting timing of impact was validated using the signal of an accelerometer on the obstacles, sampled at 1000 Hz. The detection method based on kinematic data resulted in an acceptable mean error of 1.61 ms (SD 5.82).

For onset detection of EMG activity bursts, we subtracted the averaged pattern of 5 walking trials from the averaged pattern of 5 tripping trials for each muscle. Onset was determined on these subtracted signals by means of a dynamic process model in combination with statistically optimal change detection, described by Staude & Wolf [80]. For a period of 200 ms following trip initiation, this method searched for changes in the sequence by use of the likelihood ratios over small time windows. The same method was used for onset detection of the perturbation effect on the moment generation. For each tripping trial, the averaged joint moment pattern of 5 walking trials was subtracted, and deviation of the signal following trip initiation was determined. Over the period of 50 ms following onset of this perturbation effect, the rate of change in generating the joint moments was calculated. Finally, peak values were quantified of the joint moments and of the GRF in vertical and horizontal (fore-aft) direction over stance time of the support limb for each walking and tripping trial. Differences in EMG onsets of the tripping responses were tested between muscles by paired t-tests with a Bonferoni correction. In order to test for differences between normal walking and tripping reactions, within-subject averaged (across trials) values of the variables were statistically tested in a multivariate analysis of variance (MANOVA) for repeated measures (SPSS Inc. statistical software). The level of significance selected was set at $p=0.05$.

Results

The subjects walked at a velocity of 1.61 (SD 0.15) m/s and frequency of 117 (SD 4.5) steps/min. Tripping reactions were induced on average at 39 (SD 3.8) % of the normal swing phase duration. Typically after tripping in this particular phase of the gait cycle, subjects performed an elevating strategy. Trials selected were all successful recoveries. Figure 5.2 depicts stick diagrams of a typical tripping reaction. Immediately after collision the obstructed swing leg was elevated over the obstacle while the support limb provided prolonged push-off. The duration of a stride, normally 1.03 (SD 0.04) s, was increased significantly (p<0.001) for the obstructed swing limb (1.12, SD 0.12 s) as well as for the support limb (1.26, SD 0.10 s). In the swing limb, flexion of all joints occurred to facilitate obstacle clearance and positioning of the recovery limb. In the support limb, joint rotations toward extension were seen.

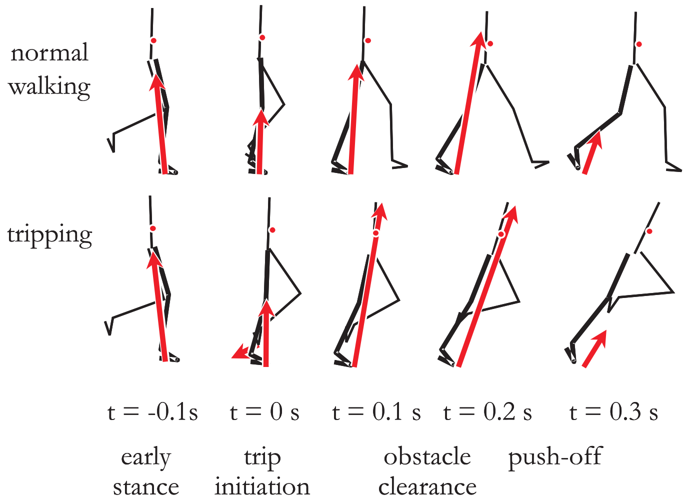

Figure 5.2: *Stick figures of a typical walking trial (upper graph) and a tripping trial (lower graph) for 5 instants of time (trip initiation at t=0) of one subject. Thin lines indicate the obstructed swing limb; thick lines depict the support limb. Body center of mass (●), foot-obstacle contact force vector at the toe and ground reaction force vectors are drawn. Center of pressure of the GRF can be positioned anteriorly of the foot because the toe marker is located at the metatarso-phalangeal joint.*

The responses and onsets seen in the support limb muscles during the primary phase of recovery after tripping are presented for a typical subject (Figure 5.3) and averaged over subjects (Table 5.1). The first muscle responses were observed in the hamstring muscles (ST and BF) followed by responses of triceps surae (GM and SO) and gluteal muscles (GL). Subsequently, quadriceps bursts were observed (VL and RF). No responses could be detected in the TA within the first 200 ms after impact with the obstacle. The onsets of the hamstrings and triceps surae muscles were significantly shorter than the onsets of the RF muscle (p<0.002). The response times of hamstring muscles were even significantly shorter than the onset of the GL muscle (p<0.001).

Table 5.1: *Onset times (and SD) in ms of muscle responses after tripping in the (contralateral) support limb and (ipsilateral) swing limb, averaged over subjects. In TA of the support limb, no responses could be detected within 200 ms after impact.*

	support limb	swing limb
gluteus maximus (GL)	79 (8)	82 (15)
biceps femoris (BF)	65 (7)	90 (18)
semitendinosus (ST)	63 (4)	86 (18)
rectus femoris (RF)	136 (38)	104 (38)
vastus lateralis (VL)	96 (22)	115 (25)
tibialis anterior (TA)		76 (8)
gastrocnemius medialis (GM)	71 (12)	119 (12)
soleus (SO)	73 (9)	101 (14)

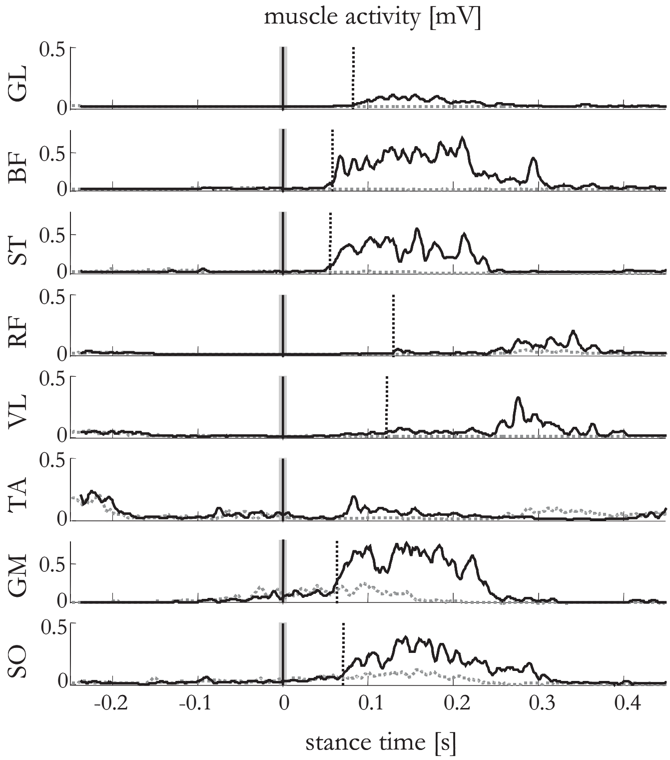

Figure 5.3: *Muscle activity patterns (mV) over stance time during normal walking (dashed) and tripping (solid), averaged over trials, for the same subject as Figure 5.2. Vertical lines indicate trip initiation (solid, with a shaded SD at t=0 s) and onset of the perturbation effect (dotted).*

The hamstring and gluteal bursts generated a hip extension moment. Figure 5.4 (typical subject) and Table 5.2 (group averages) show that whereas normally during push-off a flexion moment was observed at the hip joint, after tripping an extension moment was seen. This hip extension moment decelerated the forward angular velocity of the trunk. The sudden hamstring activity also generated a knee flexion moment, together with GM activity: whereas normally an extension moment was seen in the knee during push-off, after tripping a clear flexion moment was observed (Figure 5.4 and Table 5.2). This flexion moment decelerated the angular velocity of knee extension, and thus decelerated forward rotation of segments cranial to the knee. At the end of push-off, the flexion moment decreased and the knee was accelerated towards extension by the quadriceps activity. Finally, the rapid responses by the triceps surae muscles provided a large plantar flexion moment, needed for push-off, and a reduction of angular momentum. The peak ankle moment of the support limb was increased significantly after tripping (p<0.001; Table 5.2 and Figure 5.4), whereas the peak hip extension moment and knee flexion moment after tripping were found to be in the range of moments observed during normal walking, although at other phases in the gait cycle than push-off (Figure 5.4).

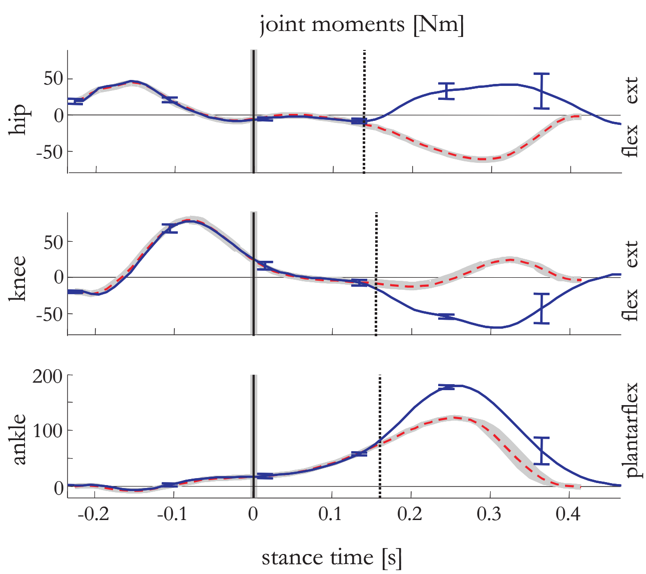

Figure 5.4: *Net joint moments for the hip, knee and ankle of the support limb for the same typical subject as Figure 5.2. Graphs over stance time for normal walking (dashed mean and shaded SD) and for tripping (solid line with error bars). Vertical lines indicate trip initiation (solid, with a shaded SD at t=0 s) and onset of the perturbation effect (dotted). Positive values in the joint moment graphs correspond with extension moment, negative values indicate a joint flexion moment.*

Table 5.2: *Mean (and SD) peak values, onsets and rate of change of joint moments and GRF for the support limb. Positive joint moments indicate extension moment; negative values indicate flexion moments. Statistical significance between conditions (*) at $p < 0.05$.*

	normal walking	tripping	
peak HIP moment (Nm)	-72 (16)	50 (26)	*
onset HIP moment (ms)		169 (24)	
rate of change HIP (Nm/s)		527 (334)	
peak KNEE moment (Nm)	45 (14)	-52 (22)	*
onset KNEE moment (ms)		167 (22)	
rate of change KNEE (Nm/s)		-519 (272)	
peak ANKLE moment (Nm)	146 (26)	202 (40)	*
onset ANKLE moment (ms)		172 (29)	
rate of change ANKLE (Nm/s)		1365 (458)	
peak GRFy (N)	201 (41)	306 (55)	*
peak GRFz (N)	927 (108)	1248 (161)	*

Joint rotations toward extension were seen in all support limb joints during push-off, but the changes in amplitudes were small (Figure 5.5). Only in the knee, a prolonged extension movement was seen. This was a result of the knee flexion moment, which decelerated the angular velocity of the knee extension. At the end of push-off, the knee was further extended by the quadriceps activity. Muscle moments generated in the joints of the support limb are reflected by the forces exerted on the floor. In all subjects, the GRF during push-off were significantly increased vertically ($p<0.001$; Table 5.2). Furthermore, the GRF were also significantly more forward directed after tripping ($p<0.001$; Table 5.2). Increased and more forward directed GRF reflect that the resulting muscle moments in the support limb will decelerate the angular momentum of the body [56]. This is reflected by a GRF vector passing the body center of mass anteriorly during push-off after tripping, as can be seen in Figure 5.3.

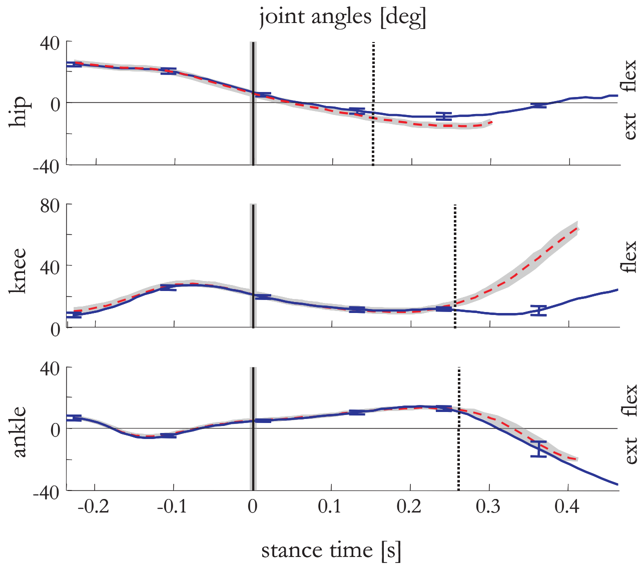

Figure 5.5: *Joint angles for the hip, knee and ankle of the support limb over stance time for the same subject as Figure 5.2, during normal walking (dashed mean and shaded SD) and for tripping (solid line with error bars). Vertical lines indicate trip initiation (solid, with a shaded SD at t=0 s) and onset of the perturbation effect (dotted). Positive values correspond with joint flexion; negative values indicate a joint extension.*

Discussion

This study investigated how early reactions of the support limb contribute to a successful recovery after tripping. The motivation for this was that, as Grabiner et al. [26] stated, "by characterizing the determinants of successful recovery, biomechanical requisites may be compared to individual performance capability, and predisposition to falling for individuals who are unable to execute specific recovery sequences may be quantitatively identified". Their study, and those of others who described tripping responses, were limited to the obstructed swing limb and to description of muscle responses and kinematics [20, 26, 27, 51, 73, 74]. Our previous study showed that the support limb could also play an important role in adequate recovery [56]. The present study investigated how such a contribution was achieved by the support limb in terms of response times and muscle moment generation, in order to provide more insight into the requirements for successful recovery after tripping. We hypothesized that the support limb would show rapid responses, high peak muscle moments and high rates of change in moment generation during push-off. If rapid and strong reactions are required, these could be limiting factors for recovery in the elderly.

Quick responses, with latencies of 60-80 ms, were seen in the muscles of the support limb, especially in the gluteal, hamstring and triceps surae muscles. These response times were about as fast as those observed in the swing limb [20, 74]. Crossed reflexes could account for these rapid reactions [3, 86]. Accounting for an electro-mechanical delay of about 100 ms [99, 100], these rapid muscle activities caused fast increases in muscle moments in the joints during the primary phase of recovery. The peak values and rate of change of these muscle moments were large. Especially a large ankle plantar flexion moment, generated by triceps surae muscles, and a hip extension moment, generated by hamstring activity, brought about the necessary push-off reaction. These net moments in the support limb were high in comparison to literature data on the capacity of human subjects. This was true especially for the moments about the ankle. The peak ankle moment of 204 Nm was substantially higher than the isometric maximum voluntary moments in a compilation of data from the literature for both young adults (maximum ~ 150 Nm) and elderly adults (maximum ~ 85 Nm) [19, 24, 34, 52, 62, 84, 95, 96]. The peak value and rate of change of the ankle moment exceed the isometric capacity of a group of elderly females [52] by factors of about 2.4 and 5.6 respectively. Of course, data derived using dynamometers cannot be compared directly with estimates of moments from kinematics. The abovementioned factors might be overestimated because no correction was made for joint angle, subject characteristics, and the fact that voluntary activation in isometric conditions is not necessarily maximal. Nevertheless the data show that high ankle moments are required during push-off, which could be a problem for the elderly who might not be able to generate such large moments.

For the hip and knee, peak values and rate of change of joint moments were less extreme. However, other striking findings appeared in these joints: the hip and knee joint moments of the support limb were directed opposite to those during push-off in normal walking. We found hip extension moment instead of hip flexion moment and knee flexion moment instead of knee extension moment. Similar reversals in joint moments were observed in slipping [9, 65]. During push-off, all joints of the support limb needed to be extended, but simultaneously, the angular momentum of the body had to be restricted. Hip and ankle extension moments meet these requirements, but a knee extension moment would increase the angular momentum. By generating a knee flexion moment, as observed, angular momentum of the body was decelerated.

Consequently, stance duration and time available for push-off was prolonged (by about 9%). The combination of a knee flexion moment and a hip extension moment, together with an increased ankle plantar flexion moment caused the ground reaction force vector to increase in magnitude and become directed more forward. This indicates an attempt to prevent a further increase or even bring about a reduction of the forward angular momentum of the body, in line with earlier findings [56]. The attempt to control forward angular momentum was observed in all subjects, although from the earlier findings it appeared that only some subjects were able to actually reduce the angular momentum to zero [56].

Contribution of the support limb in restraining the angular momentum of the body, and the trunk in particular, was suggested, but not explained before. Grabiner et al. [26] suggested that in the primary phase of recovery, trunk control might be achieved by generation of a hip extension moment by hamstring and/or gluteus maximus muscle activity, as they did not find an influence of the capacities of the paraspinal muscles (maximum effort eccentric trunk extension strength, voluntary-reaction time, automatic response latencies and activation levels) on trunk kinematics during the positioning phase. Others, who found increased activity in these muscles in the contralateral support limb after perturbation, have also described the potential effect of the hamstring and gluteal muscles on trunk extension [16, 20, 74]. The present study showed that indeed, a large hip extension moment was generated. One should bear in mind that not only the hip joint moment contributes in trunk control, but dynamic control of the angular momentum of the body involves all joints of the support limb [108].

In summary, during the primary phase of recovery after tripping, an early and pronounced increase in muscle activity was seen in the support limb. Extension reactions were observed to achieve push-off and simultaneously reduce the angular momentum of the body. A reversal of joint moments in the hip and knee, and very large ankle extension moments were required. These reactions in the support limb resulted in a push-off reaction that provided time and clearance for adequate positioning of the recovery foot and contributed to a decrease of the angular momentum during push-off. The ability of hip and ankle extensor muscles to react quickly and generate large forces in the limb is therefore an important determinant for successful recovery, at least, during the primary phase of recovery. In the next phase of recovery, when landing on the recovery limb, one would expect high net moments in the hip and knee of this limb. Preliminary results indicate that indeed large hip and knee extension moments are generated in the recovery limb during landing. Further research is required to investigate the characteristics and requirements of the landing phase. However, the inverse dynamics analysis in the present study shows that the support limb plays an important role in adequate recovery after tripping. This contribution should not be underestimated, because the more angular momentum is taken away by the support limb during push-off, the less remains to be accomplished by the recovery limb after landing. The support limb contribution can be ascribed particularly to large ankle moments (for push-off) and hip extension moments (for trunk control). These results add to the identification of factors predisposing for falls, as more falls could be expected in older adults who react less rapid and less strong than young adults. Strength training in the muscle groups responsible for push-off and trunk control may be indicated in older adults to reduce the risk of falling after a trip.

Acknowledgements

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Push-off reactions in recovery after tripping discriminate young subjects, older non-fallers, and older fallers

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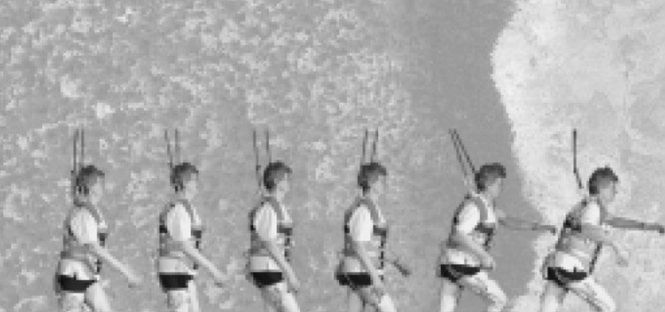

Chapter 6

Abstract

Tripping is a major cause for falls, especially in the elderly. The present study investigated whether falls in the elderly can be attributed to inadequate push-off reactions by the support limb in the recovery after a trip. Twelve young (20-34 years) and eleven older (65-72 years) men and women walked over a platform and were tripped several times over an obstacle that suddenly appeared from the floor. Kinematics and ground reactions forces of the support limb during push-off were measured of falls and successful recoveries. Young subjects did not fall. The older subjects were divided in a group of 4 non-fallers and 7 fallers. Older fallers showed insufficient reduction of the angular momentum during push-off and less proper placement of the recovery limb. This was due to a lower rate of change of moment generation in all support limb joints and a lower peak ankle moment. Onset of knee moment generation was slightly delayed in older fallers. Improvement over trials was ascribed to better positioning of the recovery limb, as no clear differences were seen in the joint moments of the support limb. In conclusion, the contribution of the support limb to prevent a fall after tripping is decreased in older adults. Lower limb strength could be an underlying factor and strength training might help to reduce fall risk.

Introduction

Identification of factors reducing the ability to prevent a fall in the elderly can be used to define intervention targets in fall prevention programs [6]. As tripping is one of the main causes for falls [2, 4, 44], several authors investigated recovery reactions after tripping in young adults [20, 23, 26, 27, 56, 57, 73, 74, 78]. Pavol and co-workers investigated recovery after tripping in a group of older adults [47, 49-52]. They found decreased lower extremity strength to increase fall-risk by limiting the ability to execute the required motor response and on the other hand to decrease fall-risk as less strong people walk slower, which makes recovery after a trip less demanding.

The essence of preventing a fall after tripping is to reduce the angular momentum, which the body acquired from impact with the obstacle. Eng et al. [20] described two strategies for recovery after tripping. An elevating strategy is observed after a perturbation in early swing and consists of an elevation of the obstructed (ipsilateral) swing limb to overtake the obstacle. A lowering strategy is seen during late swing and consists of an immediate placement of the obstructed foot on the ground, followed by a step of the contralateral limb to overtake the obstacle. For both strategies, the foot that is positioned forward after the trip is defined the recovery foot. In this paper, we focus on the elevating strategy.

Placing the recovery limb anteriorly of the body to generate force is one means to reduce the angular momentum [26, 27, 51]. In addition, the support limb (stance limb at time of tripping) plays an important role, before the recovery limb hits the ground [56, 57]. During the push-off phase (from the instant that contact of the swing foot with the obstacle ends until support limb toe-off), the support limb can contribute to recovery by generating adequate forces. This way, the support limb can provide time and clearance for proper recovery limb positioning, but can also reduce the angular momentum of the body. This can contribute to recovery success, because the more angular momentum is taken away by the support limb, the less remains to be accomplished by the recovery limb. During push-off young subjects generate fast and large ankle and hip extension moments [57]. Generating such reactions could be a problem for the elderly, since lower extremity strength, rate of force generation and reaction speed decline with age [36, 75].

The purpose of this study was to investigate 1) whether older adults react less adequate than young adults during the primary phase of recovery after tripping and 2) why some older adults fall more often than others. For this purpose, we had 12 young and 11 older subjects walk over a platform, and tripped them several times over an obstacle. Kinematics and ground reactions forces during push-off were measured. We expected older subjects to react more slowly and generate lower joint moments (relative to body mass) than young subjects during push-off. Consequently, the support limb would contribute less to reduction of the angular momentum during push-off, resulting in a higher frequency of falling.

Methods

Subjects

Twelve young and eleven older subjects voluntarily participated in this study (Table 6.1). Subjects were informed on the research procedures before they gave informed consent in accordance with the ethical standards of the declaration of Helsinki. Protocol, data collection, and part of the results of the young subjects were described previously [56, 57].

Table 6.1: *Subject characteristics; group averages (and SD).*

Group		# Subjects	Gender	Age (yr)	Height (m)	Weight (kg)
Young		12	6 ♂, 6 ♀	27.1 (4.3)	1.78 (0.07)	75.1 (8.9)
Old	Overall	11	4 ♂, 7 ♀	67.6 (2.7)	1.72 (0.11)	77.0 (9.6)
	Non-fallers	4	1 ♂, 6 ♀	67.9 (2.6)	1.71 (0.08)	74.2 (7.8)
	Fallers	7	3 ♂, 1 ♀	66.5 (3.3)	1.72 (0.15)	75.4 (11.5)

Experimental setup and protocol

Subjects were instructed to walk at a self-selected speed over a 12 by 2.5 m platform. In the platform, a force plate was mounted and 21 aluminum obstacles (15 cm height) were hidden over a total distance of 1.5 m. In about 10 out of 50 walking trials, one obstacle appeared from the ground unexpected for the subject to catch the subject's swing limb. Online kinematic data were used to calculate where and when an obstacle had to appear to initiate a trip at mid-swing. A full-body safety harness, attached to a ceiling-mounted rail, prevented subjects from falling (on the floor). The safety ropes provided enough slack for free motion, and a spring, in series with the ropes, ensured smooth catching in case of an imminent fall. For the young subjects, video data allowed for visual detection of harness assistance. For the older subjects, a force transducer (AMTI M3-1000), in series with the safety ropes, measured the force exerted on the ropes. Trials were classified as falls when the vertical force in the ropes exceeded 200 N, at which point the slack in the ropes was taken up and the compression spring (with a pretension of 200 N) started to stretch out.

Data collection and analysis

Gait kinematics were recorded using 4 Optotrak camera arrays (Northern Digital). Motions of 12 infrared-light emitting markers, bilaterally placed on joints, were tracked and 7 body segments were defined. Ground reactions forces and center of pressure of the support limb were measured with a custom-made strain gauge force plate. All data were collected and synchronized at a sample frequency of 100 Hz.

For each subject, 5 trials of normal walking were selected with complete kinematic and dynamic data. For the young subjects, 5 tripping trials at mid-swing were selected. For the older subjects, available tripping trials ranged from 1-6, as some subjects also performed a lowering strategy at mid-swing, which could not be used for these analyses.

Heel strike, toe-off and obstacle-foot contact were detected, based on kinematic data [60]. For the older subjects, this method of detecting timing of obstacle-foot contact was evaluated using the signal of an accelerometer on the obstacles, sampled at 1000 Hz. The detection method based on kinematic data resulted in an acceptable mean error of 1.61 ms (SD 5.82). Data were analyzed in the sagittal plane after smoothing with a fifth order filter [57]. To investigate the contribution of push-off by the support limb on control of the angular momentum, we calculated the external moment (M_{ext}), which equals the rate of change in the angular momentum of the entire system. M_{ext} was calculated as the sum of the moments about the body center of mass, generated by ground reaction force and obstacle-foot contact force [56]. In addition, we investigated the contribution of the support limb to gain time and clearance. Time parameters were stance, swing and double support duration. Clearance parameters were stride length, hip height and horizontal hip displacement over a stride. The latter was taken into account as stride length is dependent on actions of both the support limb and the recovery limb, but horizontal hip displacement during the recovery step can only be achieved by actions in the support limb.

Internal joint forces and moments during push-off were calculated using an inverse dynamics model. For each segment, mass, center of mass position and the moment of inertia were calculated per subject, according to Plagenhoef [61].

To minimize possible gender effects, joint moments and M_{ext} were corrected for body mass. Changes in joint moments in response to trip initiation (moment onset) were determined for each trial by means of a dynamic process model in combination with statistically optimal change detection, described by Staude & Wolf [80]. This method searched for changes in the sequence by use of the likelihood ratios over small time windows over a total period of 200 ms following trip initiation. Furthermore, the rate of change in generating the joint moments was calculated for the period of 50 ms following the moment onset and finally the peak joint moments were determined.

Based on the ability to regain balance during the trials (see results), the older subjects were divided in two subgroups: fallers and non-fallers. In order to test for statistically significant differences between young subjects, older non-fallers, and older fallers, we used multivariate analyses of variance (MANOVA) on within-subject averaged (across trials) values. This analysis was used to test for differences in support limb joint moments (onset, rate of change and peak values), in the control of the angular momentum (external moment), and in the contribution to gain time and clearance. Tukey post-hoc tests were used for evaluation of differences between the groups of young subjects, older non-fallers, and older fallers. Significance level was set at p=0.05.

Results

Tripping reactions were induced at mid-swing, corresponding to 40% of the normal swing phase duration for all subjects. We investigated reactions in which the obstructed limb was lifted over the obstacle (and became recovery limb), while the support limb provided push-off. Immediately after collision the obstructed swing leg was elevated over the obstacle while the support limb provided prolonged push-off. Figure 6.1 depicts stick diagrams of typical tripping reactions in a young and in an older subject. None of the young subjects fell, but 7 of the 11 older subjects fell the first 1 or 2 times they were tripped. Based on the results, these subjects were classified as fallers. Six of them were female. Further results describe the joint moment generation during push-off, the reduction in angular momentum during push-off and the positioning of the recovery limb for the group of young subjects, older non-fallers, and older fallers.

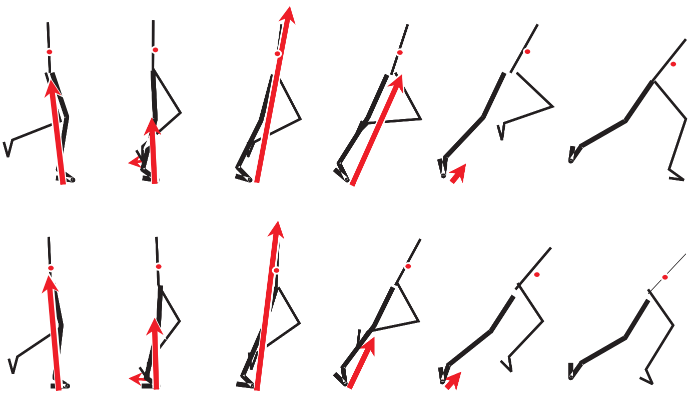

Figure 6.1: *Stick figures of a typical tripping trial for a young subject (upper graph) and an older faller (lower graph) for 6 instants of time. Thin lines indicate the obstructed swing limb; thick lines depict the support limb. Body center of mass (●), ground reaction force vectors and foot-obstacle contact force vector at the swing toe are drawn and obstacle-foot contact phase and push-off phase are indicated. An increased ground reaction force vector indicates more linear acceleration. Depending on its direction, this force indicates angular acceleration (passing the body center of mass posteriorly) or deceleration (passing the body center of mass anteriorly). Note that the period from toe-off until placement of the recovery limb is an aerial phase in the young subject, whereas these events occur almost simultaneously (double support) in the older subject.*

Joint moments

During push-off after tripping, all young and older subjects showed a hip extension moment, a knee flexion moment and an increased plantarflexion moment in the support limb. These internal moments reduce the forward angular velocity of the body, while providing extension for push-off [57]. The onset of a change in joint moment generation was slightly different between the young and the older fallers only in the knee. The rate of change of moment generation in all joints and the peak ankle moment were significantly lower for the older fallers than for the young and non-fallers (Figure 6.2). Muscle moments generated in the joints of the support limb are reflected by the forces exerted on the floor (Figure 6.1).

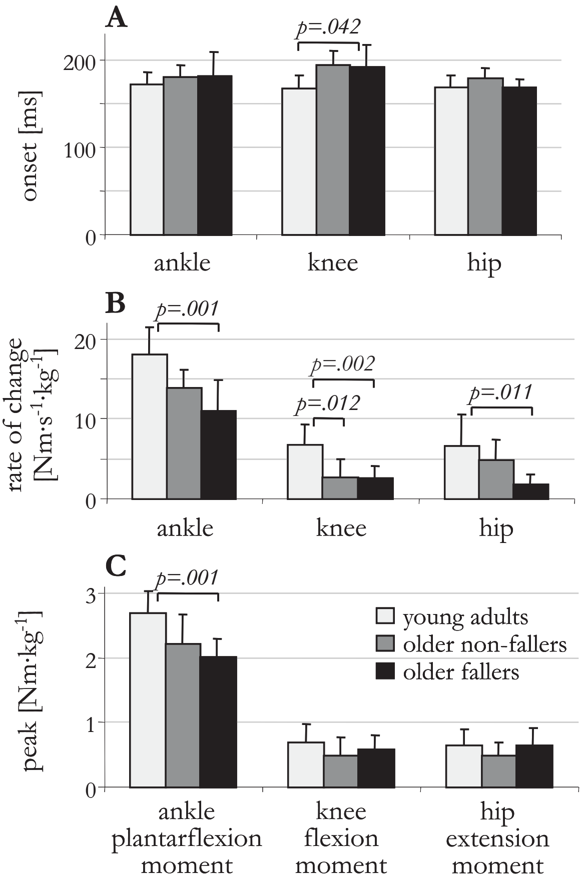

Figure 6.2: *A) moment onset (ms after initial obstacle-foot contact), B) moment rate of change ($Nm \cdot s^{-1} \cdot kg^{-1}$ over 50 ms after moment onset) and C) peak moment ($Nm \cdot kg^{-1}$) for the support limb joints during push-off. Graphs represent the internal moments: ankle plantarflexion moment, knee flexion moment, and hip extension moment. Averages over trials and subjects are plotted for the groups of young, older non-fallers, and older fallers. The error bars indicate standard deviations over the group after averaging within subjects. Significant differences between the groups are indicated with p-values.*

Control of angular momentum

Figure 6.3 depicts M_{ext} over time for 3 typical subjects: a young subject, an older non-faller and an older faller. All subjects started rotating forward during obstacle-foot contact, due to contact force and gravity. The increase in angular momentum is reflected in a positive M_{ext}. The area under the curve of M_{ext}, which equals the change in angular momentum, was similarly increased during obstacle-foot contact for all groups (Figure 6.4).

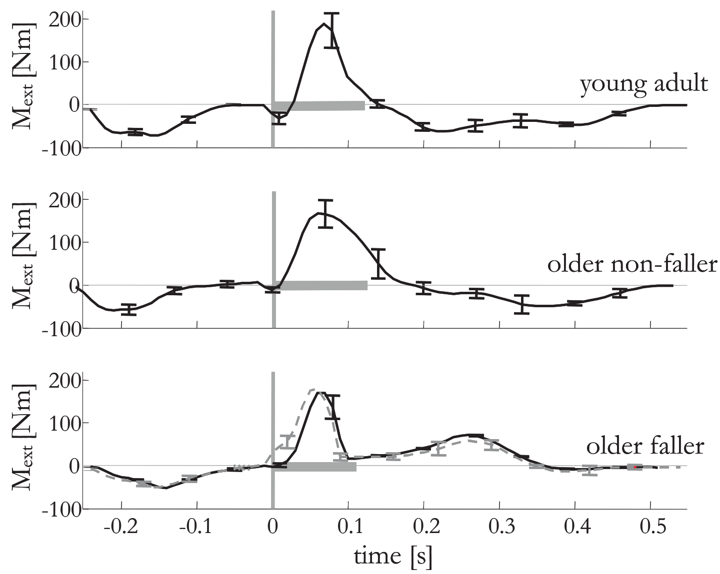

Figure 6.3: *The external moment (M_{ext} in Nm) over time for 3 typical subjects; a young subject, an older non-faller, and an older faller. Graphs are subject averages (± within subjects SD) over the primary phase of recovery: from heel strike of the support limb to landing of the recovery limb. For the older faller, the solid line represents the average of 2 successful trials and the dashed line represents the average of 2 fall trials. The vertical lines at t=0 s indicate trip initiation, followed by horizontal lines indicating obstacle-foot contact duration. Note that the change in the external moment just prior to trip initiation is due to differentiation. A positive M_{ext} reflects an increase of angular momentum (clockwise angular acceleration); a negative M_{ext} indicates a decrease of angular momentum (counterclockwise angular acceleration). The young subject is able to reduce the angular momentum during push-off, the older non-faller is able to prevent the angular momentum from increasing, but the older faller was not able to restrain the angular momentum; it continued to increase during the push-off phase (in fall trials as well as in successful recoveries).*

If subjects would not react, gravity would cause a further increase in angular momentum. Generating adequate joint moments can prevent further increase. This is reflected in a negative M_{ext}, following the positive phase (Figures 6.3 and 6.4). If the area under the curve during push-off equals the area during obstacle contact, complete recovery has been achieved. Complete recovery was seen in 4 of the young subjects. The remaining young subjects were able to prevent a further increase in forward rotation (due to gravity). None of the older subjects was able to fully reduce the angular momentum during push-off, but restraining further increase was achieved by the older non-fallers. The older fallers, however, were not able to restrain their angular momentum and their angular acceleration continued to increase during push-off (Figures 6.3 and 6.4). These older subjects had more difficulty to recover and used their harness for at least one of the tripping trials. No major differences could be found in M_{ext} or joint moments between the fall trials and the more successful recoveries within the group of older fallers (see Figure 6.3). So the fallers were less able to generate adequate moments during push-off, but this did not always result in a fall.

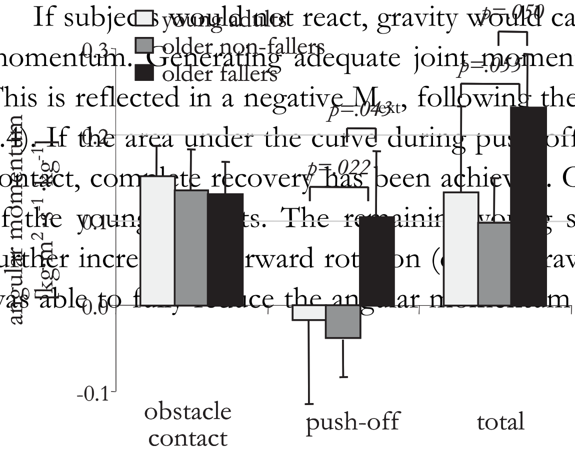

Figure 6.4: *The change in angular momentum ($kg \cdot m^2 \cdot s^{-1} \cdot kg^{-1}$, calculated as the area under the M_{ext} curve) during the obstacle-foot contact phase, the push-off phase and the sum of both phases, for the three groups. Significant differences between the groups are indicated with p-values. The angular momentum obtained during contact was equal, but whereas the young and older non-fallers were able to restrain the angular momentum during push-off (evidenced by a negative value), the angular momentum of the older fallers continued to increase during push-off (positive value). The total amount of angular momentum at landing of the recovery limb was therefore significantly greater for the older fallers.*

Time and clearance for positioning recovery limb

Young subjects walked at a speed of 1.61 (SD 0.15) $m \cdot s^{-1}$ and frequency of 117 (SD 4.5) $steps \cdot min^{-1}$. Older adults walked at the same frequency, but their averaged speed was significantly lower at 1.44 (SD 0.18) $m \cdot s^{-1}$, due to smaller strides (Figure 6.5). No significant differences were found in walking speed, stride length, or obstacle contact phase between older non-fallers and fallers.

In young subjects, push-off by the support limb also contributed to recovery by providing time and clearance for proper positioning of the recovery limb. Prolongation of the stance phase duration of the support limb was seen, followed by an aerial phase instead of a double support phase (Figures 6.1 and 6.5A). Concurrently, the swing phase duration of the recovery limb was increased. Older subjects also increased the stance time of the support limb and decreased the double support time, but did not achieve a clear aerial phase. The difference in swing phase duration of the recovery limb was not significantly different between older non-fallers and older fallers (Figure 6.5A).

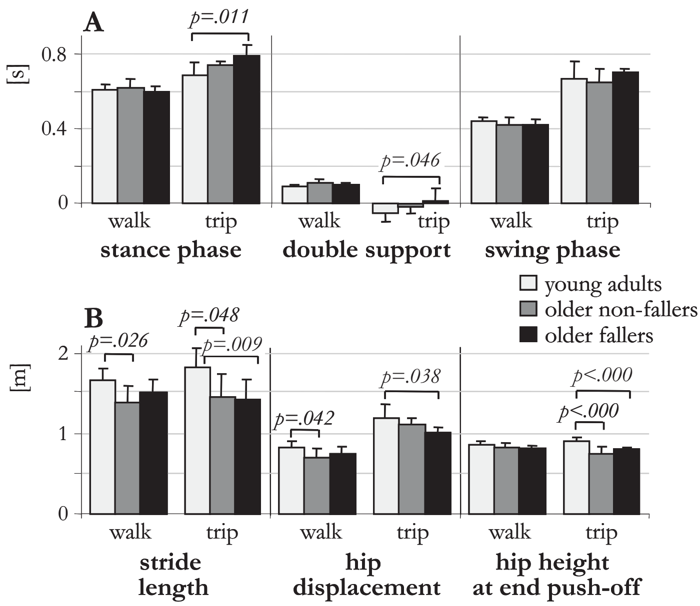

Figure 6.5: *A) Timing and B) clearance during push-off for normal walking and tripping. Averages over trials and subjects are plotted for the groups of young, older non-fallers, and older fallers. The error bars indicate standard deviation over the group after averaging within subjects. Significant differences between the groups are indicated with p-values. After a trip, young subjects showed an increased stance phase duration in the support limb, an aerial phase instead of double-support and an increased swing phase duration of the recovery limb. During the prolonged swing phase, the recovery limb was swung further forward (increased recovery stride length and horizontal hip displacement during the recovery step) and the body was elevated (increased hip height at end push-off). Older subjects showed no major differences in the timing variables compared with the young, but there was no clear aerial phase and recovery stride length, hip displacement and hip height at end push-off were less, particularly in the older fallers.*

During a prolonged swing phase, the recovery limb can be swung further forward, as seen in young subjects. Older non-fallers showed less stride length increase than young subjects, and older fallers even showed a decreased recovery stride length after tripping. Stride length is dependent on actions of both the support limb and the recovery limb, but horizontal hip displacement during the recovery step can only be achieved by actions in the support limb. Young subjects increased the horizontal bilateral hip displacement. Older non-fallers and fallers also increased their horizontal displacement of the pelvis, but to a lesser extent and not significantly different from each other. Furthermore, young subjects elevated their body at the end of the push-off phase for the aerial phase; at end push-off, the averaged hip height was about 5 cm higher after tripping compared to normal walking. The older subjects did not show an elevation of the pelvis at end push-off (Figure 6.5B).
The fallers were as able as the non-fallers in increasing the recovery swing duration and moving the pelvis forward, but they tended to be less able to increase the recovery stride length. Consequently, the fallers could not position their recovery limb in front of the body (for example, Figure 6.1). Indeed, we found in almost all fall trials that the recovery foot was positioned posterior of the pelvis at time of recovery limb placement (Figure 6.6). This makes a further reduction of the angular momentum by the recovery limb very difficult; the recovery limb has to generate forces to accelerate vertically but prevent further forward angular acceleration.

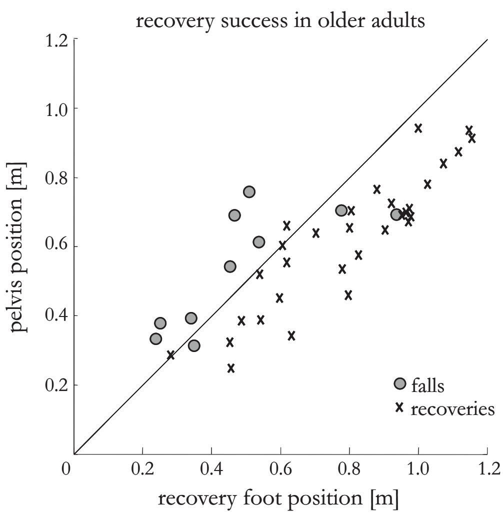

Figure 6.6: *Recovery limb positioning relative to pelvis position at landing for all individual trials of the older subjects. The units on both axes are with respect to a fixed reference point on the walkway. Successful recoveries as well as fall trials are indicated. Note that in most falls, the recovery foot was positioned posterior of the pelvis.*

Discussion

The purpose of this study was to investigate whether older adults react less adequate than young adults during the primary phase of recovery after tripping and why older fallers fall more often. The contribution of the support limb in recovery was described previously for young subjects [56, 57]. The present study showed that this contribution was considerably decreased in older subjects, and in particular in those who actually fell. Older fallers showed insufficient reduction of the angular momentum during push-off and less proper placement of the recovery limb, due to a lower rate of change of moment generation in all support limb joints and a lower peak ankle moment. Below, we will further discuss the validity of our experimental setup, the observed recovery strategies, the selected group of fit older subjects, and finally the possible implications of the present results for fall prevention.

Our experimental setup made it possible to trip subjects repeatedly at an exact point in the gait cycle. It might be argued that gait and tripping responses are altered after repeated tripping. In earlier studies with young subjects, however, we found that the normal gait kinematics were only minimally affected [60], and that the tripping trials were reproducible [56]. In the present study, the low variability of the gait kinematics and tripping responses indicates that valid experimentation with respect to tripping reactions is possible. A harness ensured safety for the falls occurring in the present study. Slight safety-rope assistance could occur during the recoveries, but only at the end of the support limb action, around recovery limb placement, and full harness assistance occurred only *after* recovery limb placement.

We tried to elicit several tripping trials at mid-swing. At this point in the gait cycle, young subjects performed an elevating strategy; the obstructed limb was lifted over the obstacle while the support limb provides push-off [20, 74]. For the elderly, we also expected an elevating strategy at mid-swing, but several older subjects performed a lowering strategy in some trials; the obstructed limb was lowered before the obstacle, and the support limb becomes the recovery limb. Several lowering strategy trials resulted in a fall as well. We did not analyze these trials in the present study, because we did not measure the push-off forces of the obstructed limb. However, these trials suggest that strategy selection might be a factor determining recovery success and needs further investigation.

A selected group of fit and healthy older subjects participated in this study. Nevertheless, the majority of the older subjects fell at least once. Higher walking velocity in fit older subjects can increase the likelihood of falling in older adults [49, 51]. However, in our study, walking velocity did not differ between non-fallers and fallers, nor did the perturbations effect (i.e. the amount of the angular momentum acquired in the obstacle-foot contact phase). Falls could therefore not be ascribed to differences in walking velocity. Most of the older fallers were women. There might be relation between age, gender and falls; we found muscle strength to be the most obvious explanation. Lower extremity strength and rate of force generation is known to decrease with age and the decrease is greater for women than for men [36, 75]. Increased age and female gender have been related to a higher fall risk [52, 76, 104, 105]. If recovery success is indeed related to lower extremity strength, it can be assumed that even more falls would occur in a less fit group of older adults.

The present study showed that recovery success after tripping is largely determined during push-off by the support limb. The more the angular momentum is reduced during push-off, the less remains to be accomplished by the recovery limb after landing. Insufficient reduction of the increased angular momentum and less proper placement of the recovery limb were seen in the group of older fallers. This was due to a lower rate of change of moment generation in all support limb joints, a lower peak ankle moment and a slight delay in knee moment onset. Although we did not measure joint moment generating capacity in the lower limbs, this appears to have been the cause for falls in some older subjects. Strength training of responsible muscle groups (i.e. hip extensors and plantarflexors for push-off [57] and possibly knee extensors to control a collapsing knee joint after landing) may therefore be indicated in older subjects to reduce the risk of falling after a trip. Of course, recovery continues after placement of recovery limb. It was seen that older fallers improved their recovery success over trials. As no clear changes were seen in the joint moments of the support limb, this suggests that the inadequate moment generation is typical for the older fallers and predisposes to a fall. It seemed that after a few trials, older fallers were able to compensate less adequate performance in the push-off phase, by better positioning of the recovery limb. Furthermore, increased recovery stride length implies more forward swing of the recovery limb during push-off, which has an additional restraining effect on the forward angular momentum. Besides push-off reactions, forward swing of the lower limbs can be an important target in fall prevention training.

Conclusion

The contribution of the support limb to prevent a fall after tripping is decreased in older adults, particularly in older fallers. Older fallers showed insufficient reduction of the angular momentum during push-off and less proper placement of the recovery limb. This was due to a lower rate of change of moment generation in all support limb joints and a lower peak ankle moment. Strength training might help to reduce fall risk.

Acknowledgements

The authors would like to thank Richard Casius and Leon Schutte for developing the data acquisition software and for help with the experiments and Petra van der Burg for helpful comments on this manuscript.

Control of support limb muscles in recovery after tripping in young and older subjects

Experimental Brain Research, conditionally accepted

Chapter 7

Abstract

Older people fall more often after tripping than young people, due to a slower development of mechanical responses. This might be due to age-related changes in muscle properties, but also to changes in motor control. The purpose of the present study was to determine whether a) timing and sequencing of muscle activation and b) the magnitude and rate of development of muscle activation in recovery after a trip differs between young and older subjects. We focused on the support limb, as it contributes to recovery after tripping by counteracting the forward angular momentum. Ten young (25 years) and seven older (68 years) men and women walked over a platform, and were tripped several times at different points in the gait cycle. Kinematics and EMG of the support limb muscles were measured. After tripping, rapid EMG responses (60-80 ms) were observed in hamstring and triceps surae muscles in both young and older subjects. A slightly increased delay (11 ms) was found only in the soleus muscle of the older subjects. The muscle activity patterns (timing and sequencing) were similar in young and older subjects, but the magnitude and rate of development of muscle activity were significantly lower in older subjects. Especially the lower rate of development of muscle activation in the support limb of older subjects is likely to reduce the rate of force generation leading to inadequate recovery responses and falls.

Introduction

Tripping over an obstacle is a balance threat that results in a fall when recovery reactions are inadequate. Indeed, tripping is found to be one of the main causes for falls and fall-related injuries, especially in the elderly population [2, 44]. Investigation of the recovery responses after tripping over an obstacle might help to identify causes of falls, particularly in elderly people with a high risk of falling.

In a previous study [58], it was shown that older subjects (in particular fallers) were less successful in their recovery than young subjects. This was attributed to a slower generation of joint moments and a lower peak ankle moment in the support limb of the older subjects. Similarly, elderly subjects were shown to be less able to recover after a sudden release from a leaning angle due to a slower development of mechanical responses [90]. A loss of muscle fibers, predominantly of type II fibers, with ageing has been demonstrated [33, 63, 91] and tendon compliance was shown to increase [66]. These changes in muscle properties would cause muscles to become slower and less strong and thus might underlie the observed age effect on the recovery from tripping. However, changes in motor control might also contribute.

Two strategies for recovery after tripping have been described, the occurrence of which depends on the time of trip initiation in the swing phase [20]. An elevating strategy is observed after a perturbation in early swing and consists of an elevation of the obstructed (ipsilateral) swing limb to overtake the obstacle. A lowering strategy is seen during late swing and consists of an immediate placement of the obstructed foot on the ground, followed by a step of the contralateral limb to overtake the obstacle. The strategies are defined on the basis of the obstructed swing limb, but a strategy-dependency was found in the support limb as well [16]. Thus, dependent on the context, appropriate muscle responses need to be selected for recovery after a trip. It is conceivable that age related changes in functioning of the basal ganglia negatively affect such response selection [5, 17]. Furthermore, studies on spinal reflexes associated with voluntary movements revealed that the delay in these reflexes increased with age [7, 35]. Changes in latencies and sequencing of responses to postural perturbations with age that were described in literature [83, 106] may be the result of these age effects on the functioning of the nervous system.

In addition, the level of muscle activation may differ between young and old subjects within the same strategy. Motor neuron excitability appears reduced in the elderly, as evidenced by lower H-reflex amplitudes as compared to young adults [71]. Pavol et al. [51] ascribed falls after tripping in a group of elderly subjects to a slower execution of the recovery strategy. In slipping, older subjects show the same phase-dependent strategies and onset latencies as young subjects, but lower levels of muscle activation during the response [85].

The purpose of the present study was to determine whether control of the support limb muscles after tripping differs between young and older subjects. We questioned whether a) timing and sequencing of muscle activation and b) the magnitude and rate of development of muscle activation in recovery after a trip is changed with age. It was hypothesized that in older adults, muscle response time would be delayed, activation sequence would be altered and magnitude and rate of development of muscle activation would be decreased. For this investigation, we had young and older subjects walk over a platform, and tripped them several times over an obstacle at different points of the gait cycle to elicit elevating as well as lowering strategies. Muscle activation patterns of the support limb muscles were measured and onsets, amplitudes and rise times (time from response onset to response peak) were compared between the young and older subjects.

Methods

Subjects

Participants in this study were 10 young subjects (5 female, 25.2 ± 4.2 years), and 7 older subjects (5 female, 68.3 ± 3.0 years). All subjects were fit and healthy and screened for having no orthopaedic, neuromuscular, cardial or visual problems. Subjects were informed about the research procedures before they gave consent in accordance with the ethical standards of the declaration of Helsinki. Protocol and data collection were described previously [57].

Experimental setup and protocol

Subjects were instructed to walk at a self-selected speed over a 12 m by 2.5 m platform and instructed by feed-back to maintain this walking velocity. In the platform, 21 aluminum obstacles (15 cm height) were hidden over a total distance of 1.5 m. During several walking trials, one of these obstacles appeared from the ground unexpectedly for the subject to catch the subject's swing limb. The obstacles appeared about 100 ms before impact, which minimized an effect of sound and sight of the appearing obstacle. Young subjects were tripped in about 15 out of 60 walking trials. Older subjects were tripped in about 7 out of 40 walking trials, as the experiment was more strenuous for these subjects. At the start of each trial, subjects did not know whether or where an obstacle would appear. Online kinematic data of toe markers were used to calculate the subject's step length and velocity. Based on these variables, the computer calculated when and where relative to the stance limb an obstacle had to appear to initiate a trip in a specific phase of the stride cycle (i.e. early or late swing). A full-body safety harness, attached to a ceiling-mounted rail, prevented subjects from falling. The safety ropes provided enough slack for free motion, and a spring in series with the ropes ensured smooth catching in case of an imminent fall. A trial was classified as a fall when the vertical force in the ropes, measured by a force transducer (AMTI M3-1000), exceeded 200 N.

Data collection and analysis

Gait kinematics were recorded using 4 arrays of 3 cameras (Optotrak, Northern Digital). Motions of 12 infrared-light emitting markers, bilaterally placed on joints, were tracked at a sample frequency of 100Hz. Furthermore, electromyograms (EMG) were recorded from the main muscles of the support limb: m biceps femoris (BF), m. semitendinosus (ST), m. rectus femoris (RF), m. vastus lateralis (VL), m. tibialis anterior (TA), m. gastrocnemius medialis (GM), and m. soleus (SO). Bipolar Ag/AgCl (Medicotest A/S) surface electrodes were attached after cleaning and gentle abrasion of the skin. Electrodes were placed over the mid muscle belly, in line with the direction of the fibers. The center-to-center electrode distance was 2.5 cm. The EMG signals were amplified 20 times (Porti-17tm, Twente Medical Systems), high-pass filtered (5 Hz), and stored on disk at a sample frequency of 1000 Hz with a 22-bit resolution. Next, the signals were whitened (fifth order) [14] to reduce the influence of tissue filtering and movement artefacts, Hilbert transformed, rectified and finally low-pass filtered (fifth order Savitzky-Golay filter). This filtering method preserves sudden activity onset without producing a phase-lag.

For each young subject, 10 trials of normal walking and 10 to 15 left-leg tripping trials were selected from all trials with complete kinematic, dynamic and EMG data. For the older subjects, who had fewer trials with complete data available, 5 walking trials and 3 to 7 tripping trials could be selected. The recovery attempts of the tripping trials were classified and grouped as elevating or lowering strategies, based on kinematics of respectively elevation or immediate placement of the obstructed swing limb. Heel strike, toe-off and obstacle-foot contact were detected, based on kinematic data [56].

For analysis of the EMG patterns, the time series of filtered and rectified EMG (in mV) of the undisturbed walking trials of each subject were averaged and subtracted from the EMG time series of the individual tripping trials. For comparison of muscle activity (timing and sequencing) among subjects and strategies, the resulting data of the responses of each muscle were normalized with respect to the maximum EMG activity during the walking trials. Onsets of the muscle responses were determined on the resulting signals by means of a dynamic process model in combination with statistically optimal change detection, described by Staude and Wolf [80]. This method searches for changes in the EMG sequence by use of the likelihood ratios over small time windows, over the first 200 ms after trip initiation. The rise times of the muscle responses were calculated as the time from onset of the response to 90% of the response peak. Furthermore, for a period of 300 ms following trip initiation, the mean amplitudes of the responses were determined over windows of 20 ms.

For statistical analysis of differences in EMG responses between young and older subjects, within-subjects averaged (across trials) parameters were tested in a multivariate analysis of variance (MANOVA) for repeated measures. For each muscle, differences in onset, rise times and amplitudes were tested and comparisons were made between both age groups. Differences in response amplitudes were tested over time windows for significance between strategies using post-hoc paired t-tests. The level of significance was set at $p=0.05$.

Results

All subjects walked at a constant velocity between walking and tripping trials. Walking velocity was not significantly different between young and older subjects (respectively 1.59 (SD 0.23) and 1.41 (SD 0.22) m/s), nor were EMG amplitudes significantly different between age-groups across different muscles. None of the young subjects fell during the experiments, but all of the older subjects fell in one or two tripping trials. Figure 7.1 presents the typical time series of the EMG responses of the support limb muscles for a young and older subject for walking and for both the elevating and lowering strategies. A clear difference in responses between the two strategies becomes apparent after about 200 ms, both in the young subjects and in the older subjects.

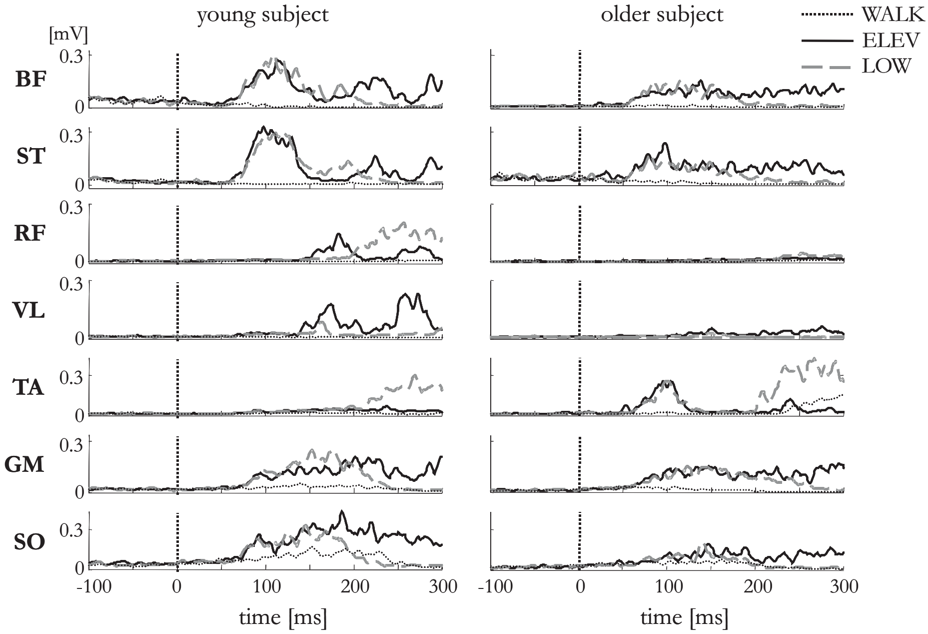

Figure 7.1: *Averaged support limb muscle activity of a young subject and an older subject, for normal walking (dotted black line), for elevating strategies (solid black lines) and for lowering (dashed gray lines) strategies. Muscles are biceps femoris (BF), semitendinosus (ST), rectus femoris (RF), vastus lateralis (VL), tibialis anterior (TA), gastrocnemius medialis (GM), and soleus (SO). Although the older subject presented here shows clear responses in the TA muscle, generally the TA responses were very small. Time series are part of the stance phase and synchronized at the averaged time of trip initiation in the stance phase. The dotted lines at t=0 ms indicate trip initiation (early stance in the elevating strategy and late stance in the lowering strategy).*

Timing and sequencing of muscle activation

Figure 7.2A shows a bar graph of the response latencies for both strategies for the young as well as for the older subjects. In both age groups, rapid responses (after about 60-80 ms) were seen in the hamstrings (BF and ST) and triceps surae muscles (GM and SO), followed by responses (after about 90-130 ms) in the quadriceps muscles (RF and VL). The responses in the TA were generally very small. Between-subjects testing revealed that a significantly increased muscle latency (11 ms) occurred only in the SO muscle of the older subjects compared to the SO latencies of young subjects. Onset times of the responses in the support limb muscles were independent of strategy and there was no significant interaction between strategy and age.

With similar onsets of the muscle activities between young and older adults, the sequencing of muscle activation appeared to be unaltered with age. Figure 7.3 depicts the mean EMG amplitudes over time of selected support limb muscles. Graphs of ST and SO were similar to those of BF and GM, respectively, and are therefore not represented. A significant interaction between strategies and time was found, indicating that indeed support limb responses are strategy dependent. The average time of divergence of amplitudes, as revealed by post-hoc t-tests (Figure 7.3), was not different between age groups: 203 (SD 41) ms in young subjects and 209 (SD 47) ms in older subjects. In the elevating strategy, the hamstrings and triceps surae muscles stayed activated, leading to a prolongation of the push-off while the obstructed swing limb was placed forward, and the VL muscle was activated which resulted in knee extension. In the lowering strategy, the hamstrings and triceps surae muscles were deactivated and the RF muscle was activated, resulting in knee extension. Furthermore, a late TA activity was seen in the lowering strategy.

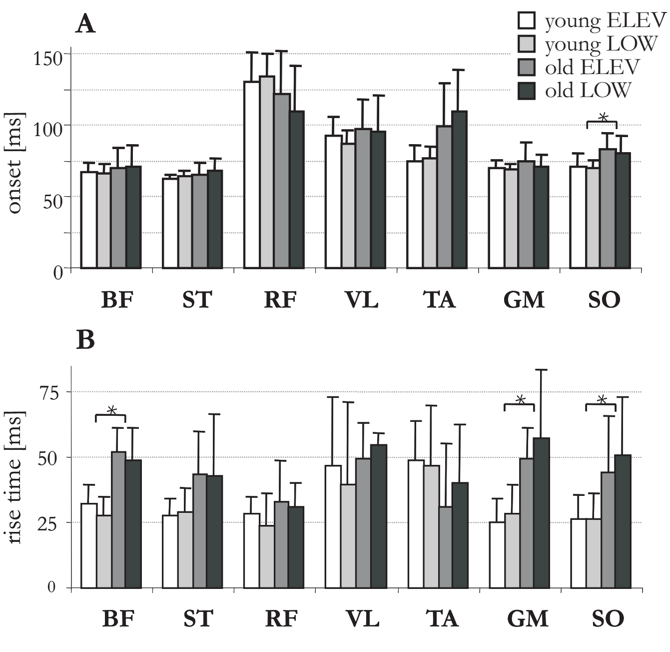

Figure 7.2: *A) Onsets of EMG activity (with SD) in the support limb muscles after trip initiation for both strategies (elevating and lowering) and age groups (young and older subjects). B) Rise times of the EMG activity (from onset till 90% of the response peak). Muscle names are the same as in Figure 7.1. Statistically significant differences (p<0.05) are indicated with *.*

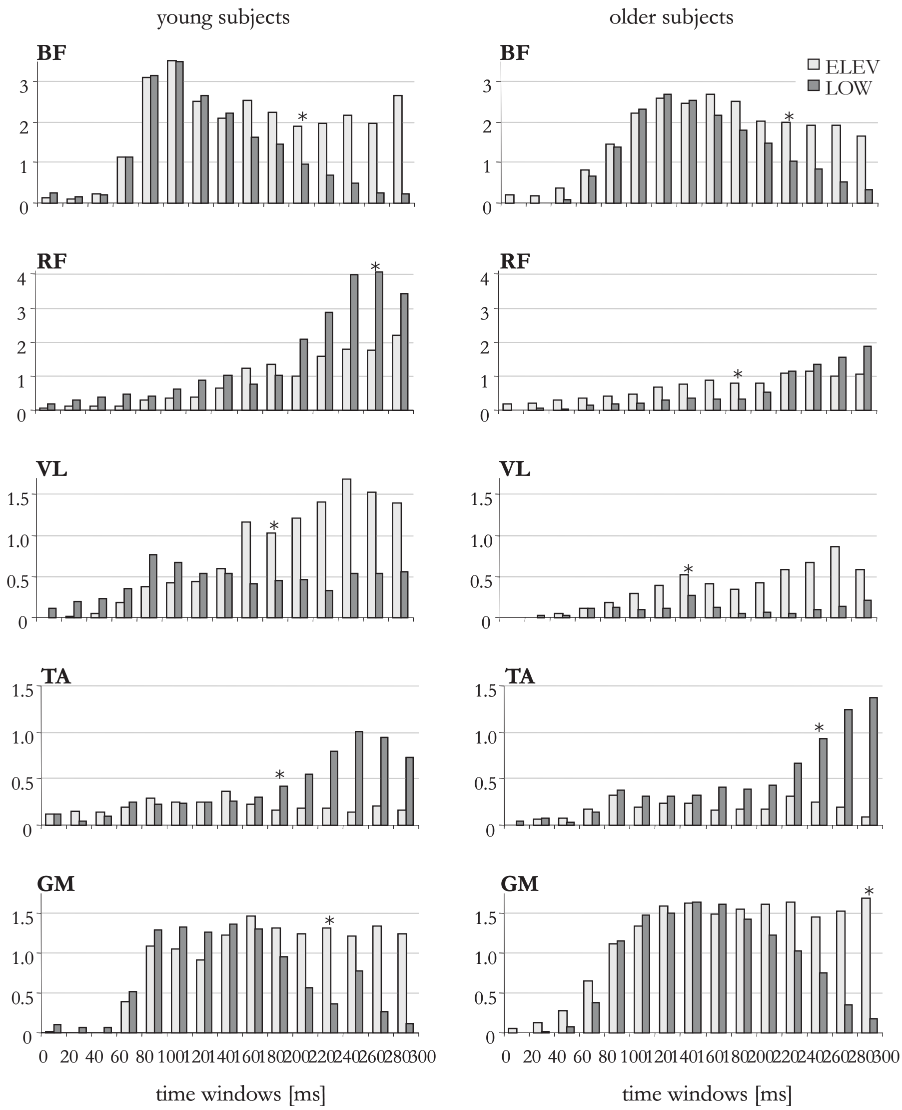

Figure 7.3: *Mean amplitudes of adaptations in muscle activity over windows of 20 ms for a total period of 300 ms following trip initiation for the support limb muscles. Muscle names are the same as in Figure 7.1. Note that the magnitude (after subtraction of normal walking activity and normalization for the maximum activity during normal walking) of the responses is larger in BF and RF than in VL, TA and GM. Time windows where amplitudes start to differ significantly between strategies are indicated by *.*

Magnitude and rate of development of muscle activation

A significant interaction effect of strategy and age on EMG amplitudes was found. No interaction effect of strategy, time, and age was found, indicating that the difference between strategies in amplitudes over time was the same for young and older subjects. From Figure 7.3 it can be deduced that the difference in strategies between young and older subjects was primarily due to a difference in the (relative) magnitude of the EMG amplitudes in most time intervals. This, in turn, appeared to be due to a slower increase in activity in the older subjects. The rise times of the EMG amplitudes were significantly longer for older subjects in the BF, GM, and SO muscles (Figure 7.2B). Note a trend towards an increased rise time in the other muscles, except for the small and variable responses in TA. Rise times of the responses in the support limb muscles were independent of strategy and there was no significant interaction between strategy and age.

Discussion

The purpose of the present study was to determine whether a) timing and sequencing of muscle activation or b) the magnitude and rate of development of muscle activation in recovery after a trip differs between young and older subjects. Both aspects of the control of the responses after a trip could be affected in older subjects, resulting in inadequate recovery after tripping. The rate at which the recovery limb is activated was addressed by other authors [20, 74]. One should bear in mind, however, that the support limb contributes to accelerating the recovery limb relatively to the upper body. The support limb provides time and clearance for proper positioning of the recovery limb [56]. Furthermore, we focused on the support limb, as this limb contributes to recovery after tripping by counteracting the forward angular momentum during push-off [56]. Young subjects achieved an adequate push-off in the support limb by fast and large moment generation through rapid responses in the hamstring and triceps surae muscles [57]. Older subjects (in particular fallers) were less successful in their recovery, mainly because they had lower rates of moment generation in the support limb joints than young subjects [58]. Given that the moments generated during tripping are high in the support limb, we focused on the large lower limb muscles. In the recovery reactions of older subjects, multiple steps were observed, as reported in other studies [37, 41]. Multiple steps in the elderly are likely to be an effect of a less effective response early on; an initial (inadequate) recovery step requires further reduction of the remaining angular momentum, for which multiple steps are inevitable.

During the experiments, subjects were tripped repeatedly at specific times of the gait cycle to elicit elevating as well as lowering strategies. The number of trips was somewhat higher in the young than in the older subjects, because the experiment was more strenuous for the older subjects. Due to repeated tripping, anticipatory changes in walking pattern might have occurred. It has been established earlier that the normal gait kinematics are only minimally affected by anticipation in young subjects [54]. In addition, although some anticipatory increase in muscle activity could occur in young and older subjects, this effect was only minimal when compared to the magnitude of tripping responses [59]. This indicates that our setup allows for ecologically valid experimentation on tripping reactions.

It should be noted that the number of subjects was small and that both males and females participated in this study. Despite the small groups of both sexes, results were significant. Gender differences in balance responses have been reported in older adults [103] and in our previous study, we also found older women more likely to fall than older men [58]. There is reason to believe that these gender differences are due to differences in muscle strength [36, 75], rather than to differences in muscle activity.

Timing and sequencing of muscle activation

We found rapid EMG responses after tripping in both age groups, which were qualitatively (in terms of timing and sequencing) similar between young and older subjects. The responses in the support limb muscles of the older subjects were not delayed compared to the young subjects, except for a slightly increased latency in the SO muscle. The patterns of muscle activity became different between strategies at 200 ms after trip initiation in both groups. It can be concluded that the healthy older subjects in our study had no difficulty with rapidly selecting the same responses as young subjects.

The most pronounced early muscle responses were observed in the BF, ST and GM muscles. These responses (with latencies of 60-80 ms) are non-specific but highly functional as they provide the hip and ankle extension moments and knee flexion moment required for successful recovery [57]. The latencies of these responses suggest that they are oligo-synaptic and highly automated, which may account for the relative robustness of their organization to the effects of ageing.

It was found that most elevating strategies were performed after a trip initiated in early swing, whereas most lowering strategies were performed when the trip was initiated in late swing, according to the literature [20, 74]. Still, in young as well as in older subjects, around mid-swing either of these strategies could be elicited. After tripping on a treadmill the transition between strategies was more distinct [74], which might be explained by lower variation of stride length and duration in treadmill walking [18]. However, the occurrence of both strategies in the same part of the gait cycle overground suggests that strategy selection is not heavily constrained, i.e. either one could be adequate. Moreover, the initial responses, which presumably are automated, provide a certain amount of time for the selection of strategy-specific responses [74]. Consequently, this strategy selection process may not differentiate the young subjects from the healthy and fit older subjects studied here. It is, however, conceivable that in a more frail population strategy selection is negatively affected by changes in functioning of the basal ganglia [5, 17].

Magnitude and rate of development of muscle activation

Initial, non-specific muscle activity increased more slowly and reached lower normalized amplitudes in the older subjects than in the young subjects. It should be noted that the amplitudes of the EMG signals were normalized with respect to the maximum EMG activity during normal walking. In spite of the lower (but not statistically significant lower) walking velocity in the older subjects, the group averaged absolute EMG amplitudes were not different between young and older subjects. Hence, we felt that is was safe to compare normalized EMG amplitudes between the groups. The rate of development of EMG activity (rise time) is independent of the normalization procedure, and is clearly lower in older subjects than in young subjects (Figures 7.2B and 7.3). The non-specific activity (during the first 200 ms following trip initiation, as found in this study) in the hamstring and triceps surae muscles in the support limb helps to restrain the angular momentum of the body, while providing extension for push-off [57]. Both are beneficial to recovery regardless of strategy. Among the older subjects several falls occurred, which are likely due to a limitation in this recovery mechanism, as older fallers showed a slower generation of joint moments and a lower peak ankle moment in the support limb than older non-fallers [58].

Probably, this is partly due to a deterioration in muscle contraction mechanisms with age [87], which can be the consequence of a range of factors such as loss of type II muscle fibers [33, 63, 91] or tendon compliance [66]. The present data, however, indicate that age related reductions in the (rate of) muscle activation might contribute to the reduced (rate of) moment generation. It is conceivable that the increased rise time in older adults reflects that the elderly subjects increased muscle activation up to a higher level to compensate for a decreased muscle capacity. However, if so, the required moment would be reached too late. Moreover, peak normalized and absolute EMG amplitudes are in fact lower in the older adults, strongly suggesting that the muscle activation is reduced and that this contributes to the lack of moment generation. The relative contribution of a decline in muscle activation with age compared to changes in muscle/tendon properties on the (rate of) force generation would require further research, for example by means of model studies.

The strategy-specific responses (after 200 ms following trip initiation, as found in this study) can yield lengthening or shortening of the push-off. In the elevating strategy, a prolonged push-off, brought about by continued hamstrings and triceps surae activity, can help to further restrain the angular momentum; moreover, it can help to accelerate the pelvis upward and forward to gain time and clearance to swing the obstructed limb forward as far as possible [56]. In the lowering strategy, a forward acceleration is not beneficial, as this would hamper immediate placement of the obstructed foot. A shorter push-off with less acceleration is required, which moreover allows making a quick step forward with the support limb for further recovery. Similarly, Dietz et al. [16] found perturbation dependent prolongations of the stance phase, if the perturbation occurred in early swing. If applied in late swing, the length of the stance phase was independent of the perturbation duration. In our experiments, we also found augmentation of the knee extension in the elevating strategy [56].

Practical implications

The present study showed that especially the rate of increase of muscle activation during recovery reactions after tripping is reduced in older subjects. This will reduce the rate of force generation in recovery after tripping, which in turn could lead to falls. It has been described that strength training can increase muscle strength in older adults [67, 77]. Importantly, these training effects are ascribed in part to neural adaptation [21, 25, 28, 63]. However, Porter et al. [63]

question the generalizability of such neural training effects across tasks. In addition, Scaglioni et al. [71] found increased voluntary activation of the plantar flexors in elderly after strength training but no increase in motor neuron excitability as evidenced by H-reflex amplitudes. It is therefore questionable whether the control of responses can be trained. Positive effects of strength training interventions [15, 69] could be due to effects on muscle properties, which may compensate for a loss in excitability.

Concluding remarks

In the control of muscle responses after tripping, the timing and sequencing of muscle responses seems to be robust to the effects of aging, whereas the magnitude and rate of development of muscle activation declines with age. These findings are in line with conclusions of other perturbation experiments which found that not response delay, but rather differences in levels of muscle activation caused an age-related decline of balance recovery [85, 88]. In particular, in our study the rate of development of muscle activation was found to be lower in the support limb of older subjects. This can contribute to the reduction of the rate of force generation in recovery responses of older adults, providing better insight why older people fall more frequently after a trip.

Acknowledgements

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Epilogue

Chapter 8

Recovery from a trip: a step ahead

The aims of this thesis were to obtain insight into the requirements for a successful recovery reaction after tripping – in particular the mechanics and control of the support limb - and to understand why older people sometimes fail to meet these requirements. The investigations of this thesis addressed the following topics: firstly the validity of the experimental setup for tripping experiments by checking for changes in walking pattern after forewarning of a possible trip, secondly the role of the support limb in recovery after tripping in young subjects, and lastly the changes with age on the mechanics and the underlying control of the support limb reactions. In this epilogue, the results and conclusions from the studies performed will be briefly summarized and revisited. Furthermore, the limitations of the present thesis are discussed, and recommendations are given for future research and for fall prevention programs.

Validity of experimental setup: changes in walking patterns after forewarning of a trip

For investigation of recovery reactions after tripping, an experimental setup was required in which tripping could be provoked. Several previous studies investigated tripping in a laboratory setting. Young adults have been tripped during over-ground walking over a suddenly appearing obstacle [20, 27], or by a rope around the ankle, blocking the swing phase [78]. Pavol et al. [49, 51, 52] were able to trip older subjects trip truly unexpectedly during over-ground walking by using an obstacle appearing from the ground. However, these experiments were necessarily limited to a single trip attempt. Young adults have also been tripped during treadmill walking by blocking their swing limb with an obstacle [72-74], or by a rope around the ankle [23]. In contrast to over-ground walking, treadmill walking has the advantage that it enables the experimenters to trip their subjects at a pre-determined point in the swing phase and multiple steps after tripping can be measured. Furthermore, it allows for a large number of “catch” trials in between actual tripping trials. However, experiments on a treadmill have the disadvantage of posing a velocity constraint on the recovery reaction and generally, exact ground reaction forces cannot be measured, which limits analysis of the mechanics. For the present thesis, an experimental setup was developed that allowed subjects to walk over-ground at a self-selected speed, while trips could be elicited repeatedly at a predetermined instant in the gait-cycle (computer controlled, based on online kinematic data). Moreover, kinematics, ground reaction forces of the support limb and muscle activity could be and were measured. Hence, mechanics as well as control of tripping reactions could be fully investigated in a controlled manner without undue limitations.

When conducting tripping experiments, subjects have to be informed about the purpose of the study for ethical reasons. Consequently, changes in the walking pattern could affect recovery reactions, causing them to be different from recovery reactions in real life. Tripping responses in an experimental setup are difficult to compare with responses in real life, but anticipatory behavior in the walking pattern can be investigated. Some of the studies on tripping described above have mentioned the possibility of anticipatory behavior in tripping experiments, but comparisons of walking pattern between normal and test walking are limited in number and scope [20, 27, 50, 72]. In this thesis, two studies were performed to investigate the ecological validity of trips elicited in our experimental setup (Chapters 2 and 3), by comparing normal walking and forewarned walking patterns. Small changes in kinematics and muscle activity patterns after forewarning were revealed. These small changes were similar in young and older subjects, even though older people are known to walk more variably [36, 91]. Both young and older subjects increased their foot clearance during mid-swing and their muscle activity after forewarning of a possible trip. The observed changes were sufficiently systematic to be statistically significant, but so small in magnitude that they were not expected to alter the probability of tripping or the recovery responses after tripping. Furthermore, a low variability of the recovery responses (see within-subjects standard deviations in Chapters 4, 5 and 6) indicated high reproducibility of the recovery reactions. Valid experimentation with respect to recovery after tripping is therefore possible in both young and older subjects. In the experiments presented in this thesis, actual tripping trials were alternated with three to five "catch" trials, the number depending on the experimenter's impression of the subject's walking pattern.

The contribution of the support limb in the recovery after tripping

The main purpose of the recovery reaction after tripping is to arrest the angular momentum, which the body obtains from impact with the obstacle. An inadequate reaction will lead to a fall. Based on results of biomechanical modeling [22], it has been suggested that trunk control after a trip is highly dependent on adequate positioning of the recovery limb; for successful recovery the recovery limb should be placed anteriorly of the hips. When properly placed, the recovery limb can generate a force and moment that counteract the body angular momentum [27]. The ability to move the recovery limb forward fast enough might thus be a limiting factor for successful balance recovery [78]. However, the contralateral support limb can help to gain time and clearance by elevating the body during push-off. In addition, the support limb has been suggested to have the potency to reduce the angular momentum of the trunk [16, 20, 74].

It has been suggested that the support limb contributes to body elevation [20] and trunk control [26], but to our knowledge, these contribution of the support limb to recovery after tripping had not yet been quantified. In our series of experiments, the contribution of the support limb was quantified in young subjects (chapter 4) and it was explained how this contribution was achieved (Chapter 5). The changes in the angular momentum of the body after tripping were estimated from the external moment (M_{ext}). It was shown that an increased and forward directed push-off by the support limb contributes to recovery by (a) providing time and clearance for proper positioning of the recovery limb (linear acceleration), and (b) restraining the angular momentum of the body during push-off (angular deceleration). All subjects were able to provide time and clearance for proper positioning of the recovery limb during push-off by the support limb. Furthermore, although not all subjects were able to reduce the angular momentum completely to zero during push-off, almost all showed a clear attempt to restrain the angular momentum (Chapter 4). To achieve this, rapid responses (60-80 ms) in triceps surae and hamstring muscles of the support limb generated a large ankle plantar flexion moment, a knee flexion moment and a hip extension moment. For the hip and knee, these moments have a sign opposite to that during normal walking (Chapter 5). Similar reversals in joint moments were observed in slipping [9, 65]. Obviously, biarticular muscles like the hamstrings and gastrocnemius muscle play an important role, as dynamic control of the angular momentum of the body involves all joints of the support limb [108].

The present thesis showed that the support limb plays a major role after tripping. Of course, this does not mean that the recovery limb is not important, as that limb has to be positioned adequately, and should further establish balance after landing [27]. Preliminary results of our experiments showed that joint moments in the recovery limb are fairly small during the positioning phase, but large hip and knee extension moments are reached during landing, shortly after this leg contacts the ground (Figure 8.1 and [55]). However, one should bear in mind that the more angular momentum is reduced by the support limb, the less remains to be accomplished by the recovery limb. In addition, the more time and clearance is provided by the support limb to swing the recovery limb forward, the more appropriate the recovery limb can be positioned (i.e. anteriorly of the body) and the easier that limb can reduce the remaining angular momentum.

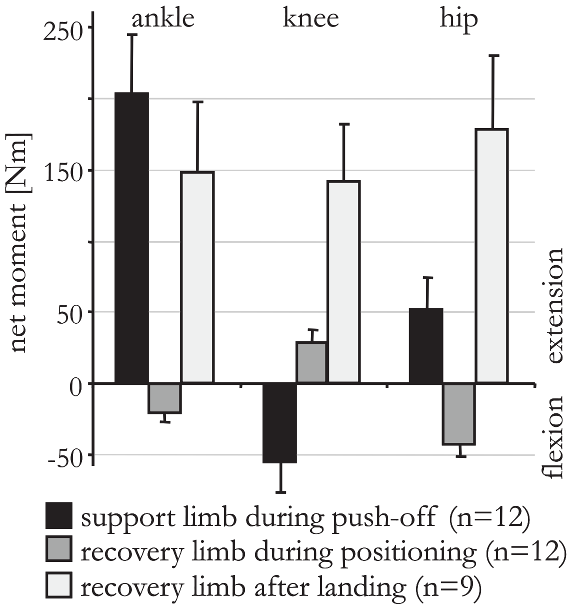

Figure 8.1: *Preliminary results on peak joint moments (and SD) in regaining balance using the elevating strategy after a trip in the ankle, knee and hip joints of both limbs of young subjects.*

Limitations in mechanics and control of recovery reaction of older subjects

Many studies have shown that muscle strength as well as the rate of force rise is lower in older subjects as compared to young subjects [e.g. 24, 62, 89]. Based on a compilation of data from the literature, Figure 8.2 shows isometric and isokinetic voluntary maximum ankle moments in young and older subjects [19, 24, 34, 52, 62, 84, 95, 96]. Given these data, it was expected that fast and strong moment generation in the support limb, as observed in the young subjects after tripping, would not be feasible for older subjects. In the study described in Chapter 6, this hypothesis was tested. It was found that older subjects used similar strategies to regain balance as young adults. Onsets of moment generation did not show substantial differences. However, the peak ankle moments and the rate of rise of ankle, knee, and hip moments (all normalized for body mass) differed between young adults, older non-fallers, and older fallers.

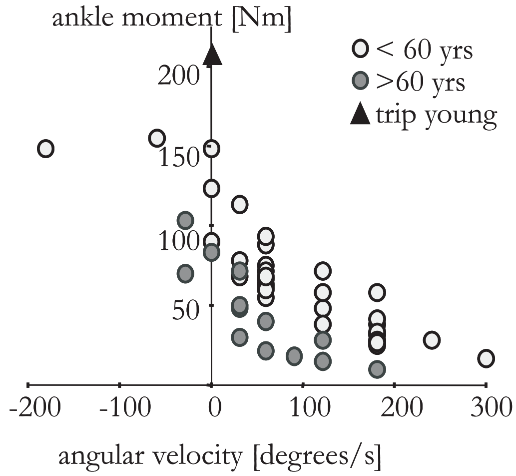

Figure 8.2: *Maximum ankle extension moments as a function of ankle extension velocity from a compilation of published sources for subjects under 60 and over 60 years of age and the maximum moments found for young subjects during tripping in the present thesis.*

Changes in muscle properties with age might underlie the deterioration of recovery reactions with age. Loss of muscle fibers, predominantly of type II fibers, has been reported [33, 63, 91] and tendon compliance was shown to increase [66]. These changes cause a decline in muscle force and rate of force development and thus might cause the slower development of moments during recovery from tripping in the older subjects. However, these age effects on recovery responses might also be caused by changes in motor control, e.g. by a decline in motor neuron excitability [63, 71]. In Chapter 7, it was shown by means of EMG that the timing and sequencing of support leg muscle responses after tripping is robust to the effects of aging, whereas the magnitude and rate of development of the muscle activation declines with age. Response patterns were qualitatively similar in young and older subjects and increased muscle latencies were found only for the soleus muscles of older subjects and were negligible in magnitude (11 ms). Similarly, studies on postural responses showed that response latencies of older subjects increase by about 10 ms. Such small increases in delay are considered not sufficient to explain the impaired balance recovery of older subjects [90, 106]. In slipping, older subjects showed the same response latencies and sequences as young subjects, but lower levels of muscle activation [85]. The results from the present thesis on tripping also showed that the levels of support limb muscle activation were reduced in older subjects. In particular, the rate of development of muscle activation was found to be lower in the older subjects. In addition to changes in muscle properties, these changes in control will reduce the rate of force generation, which can hamper the recovery mechanism and lead to a fall.

Obstacles to be overtaken

The results of this thesis have contributed to a better understanding of the mechanics and control of recovery after tripping in both young and older subjects. The studies described had some limitations. Nevertheless, the information obtained has implications for fall prevention programs.

Experimental limitations and recommendations

For the mechanical analyses in this thesis, an inverse dynamic model was used. The same model was used for young and older subjects, although the anthropometrical characteristics are known to differ between young and elderly people [53, 64]. However, the use of age adjusted anthropometical models is not likely to have large effects on the comparisons made in this thesis. Furthermore, the estimation of the angular momentum of the body was based on calculation of the external moment (M_{ext}), which equals the rate of change of the angular momentum of the body. Calculation of the angular momentum directly from the kinematic data was not deemed to be very accurate, because the angular momentum of arm segments, which made vigorous flexion and endorotation, could not be determined. For this thesis, an optimization method in the calculation of M_{ext} corrected for the arm movements. As the focus was on the lower limbs, movements of the arms were not further taken into account. Nevertheless, arm movements might contribute to balance control, but they could also serve for reaching for external supports or bracing in preparation for a fall [38, 42]. Arm elevation has also been reported to be different between young and older subjects in slipping [85]. Further investigation of arm movements, in combination with full-body models could help to understand and quantify their role in balance control.

Between-subject comparisons made by Pavol et al. [49, 51] indicate that walking speed could be a determinant of trip outcome. A higher walking speed increases the perturbation effect (i.e. the amount of the angular momentum acquired in the obstacle-foot contact phase), which makes recovery after a trip more demanding. In the experiments conducted in this thesis, walking speed was not controlled, and the effect of different walking velocities on recovery reactions was not investigated. However, walking speed did not differ between the non-fallers and fallers (Chapter 6). A model study indicated that the ability of older subjects to recover from a trip successfully does not so much improve by reducing walking velocity [97]. Yet, a faster initiation of the recovery reaction was predicted to improve the recovery success [97], pointing to the importance of the ability to react rapidly and strongly to prevent a fall. The disparity on the effect of walking speed on trip outcome needs might be solved by within-subject comparisons in future research.

All tripping studies in this thesis focused on the push-off by the support limb in recovery after tripping. It has been shown that the contribution of the support limb during push-off affects the outcome of a trip. However, even when the push-off limb is not able to generate an adequate push-off reaction, a fall could still be prevented when the recovery limb is positioned adequately. Preliminary results of our experiments on young subjects showed that joint moments in the recovery limb are fairly small during the positioning phase (Figure 8.1). During landing, however, large hip and knee extension moments are required (Figure 8.1), to accelerate vertically, but prevent further forward angular acceleration. It could be expected that older subjects may have more difficulty to generate the required muscle moments after landing. Especially, when the recovery limb is not properly placed anteriorly of the body (as observed and described in Chapter 6), the forces required to decelerate the angular momentum will be large and hence difficult to attain. When the angular momentum cannot be restrained during the push-off and landing phase, multi-step strategies will be required [23]. Indeed, older subjects often require more than one step to recover from a perturbation [41, 50, 90]. Obviously, the requirements for landing and the need to make multiple steps are related to the success in the first stage of the recovery. After all, the more angular momentum is taken away by the support limb during push-off, the less remains to be accomplished by the recovery limb after landing. More insight is still needed into the characteristics of the landing phase and requirements to make additional steps to complete a successful recovery after tripping.

Recovery reactions after tripping have been shown to decline with age. Slower moment generation and a lower ankle peak moment were found in the recovery reactions of older subjects (Chapter 6). This was partly explained by a decline in the rate of development and amount of muscle activation (Chapter 7). Muscle strength, however, also plays an important role in the deterioration of recovery reactions with age. In the studies presented in this thesis, we did not measure lower limb strength. A selected group of fit and healthy older subjects participated in the experiments. Yet, the majority of the older subjects fell at least once after being tripped experimentally. One would expect even more falls to occur in a less fit group of older adults, as they are assumed to be less able to generate the required muscle moments. Decreased lower extremity strength has indeed been described to increase the likelihood of falling [52]. Paradoxically, older subjects with higher strength of their leg muscles also ran a higher risk of incurring a fall, because they walk faster [51, 52]. In the studies presented in this thesis, we did not measure joint moment generating capacity of the lower limbs. Hence, for future research, it would be interesting to measure the older subjects' muscle properties (e.g. force generating capacities, muscle mass, tendon compliance) and to search for a relation between these properties and the recovery after tripping. Of course, walking velocity should be controlled in these comparisons. Such measurements will increase our knowledge on the requirements to recover from a trip successfully and might also lead to identification of an individual's fall risk. Furthermore, this will allow for analysis and evaluation of the effects of fall prevention programs (e.g. strength training) on both muscular properties and recovery success after tripping.

Practical implications for falls in the elderly

The present thesis showed that the ability to generate rapid and large muscle moments is important for successful recovery after tripping. The low peak ankle moments after a trip in older fallers, as compared to non-fallers and young adults, suggest that muscle strength may be a limiting factor (Chapter 6). The data also showed that a reduced rate of activation in part underlies the unsuccessful responses observed in some of the older subjects, which suggests that there might also be limitations in control (Chapter 7). It has been reported that strength training can increase muscle strength in older people [67, 77], and these training effects are ascribed in part to neural adaptation [21, 25, 28, 63]. An important question is, therefore, whether training effects transfer from the strength training task to other tasks [63]. For example, whereas voluntary activation in isolated plantar flexion has been shown to increase with strength training in older subjects, no increase was found in motor neuron excitability as evidenced by H-reflex amplitudes [71]. This could imply that the control of reflexive responses, such as the recovery responses after tripping, are not trained. On the other hand, the effect of training on muscle properties might compensate for the loss of excitability in older subjects. After all, positive effects of strength training interventions have been reported [15, 69].

The effect of strength training on the performance of functional tasks such as recovery from a trip requires further investigation. Training could be targeted at specific muscles groups that play an important role in recovery after tripping as derived from this thesis. Following this line of reasoning, especially the ankle and hip extensor musculature should be trained, considering the high demands on these muscles in the support limb for successful recovery after tripping. In addition, high demands are expected on the knee extensor muscles after landing, which suggest that knee extensor strength also should be trained. Interestingly, we found that older fallers improved their recovery success over trials (Chapter 6). Perhaps training at a more functional level could be helpful too, e.g. by applying repeated perturbations such as tripping. Obviously, further research of the effects of different interventions on muscle properties and on the ability to recover from a trip is indicated. Training effects, even if they do not improve the recovery reaction in tripping, are likely to have a positive effect on general health and on the quality of life in elderly people.

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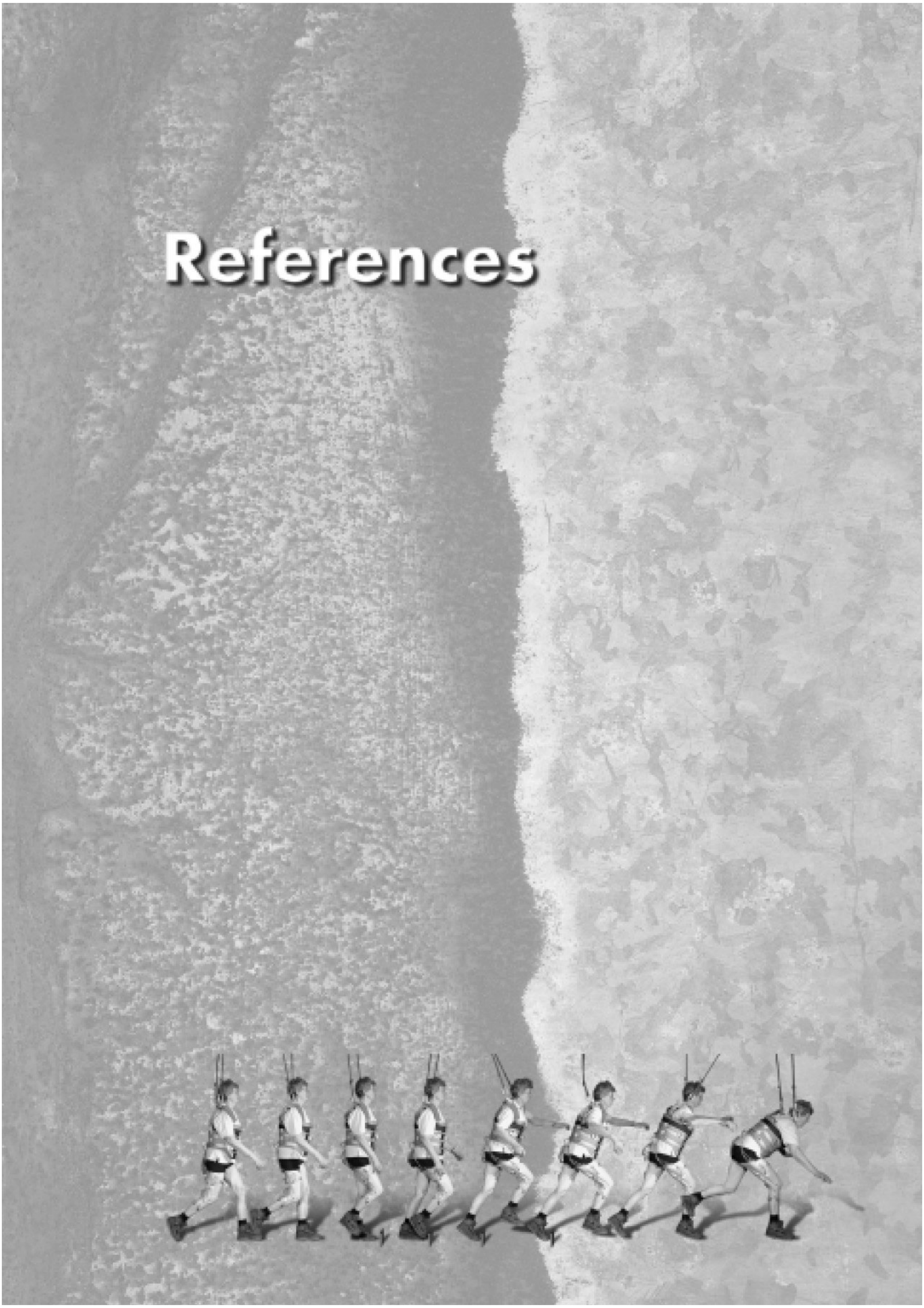

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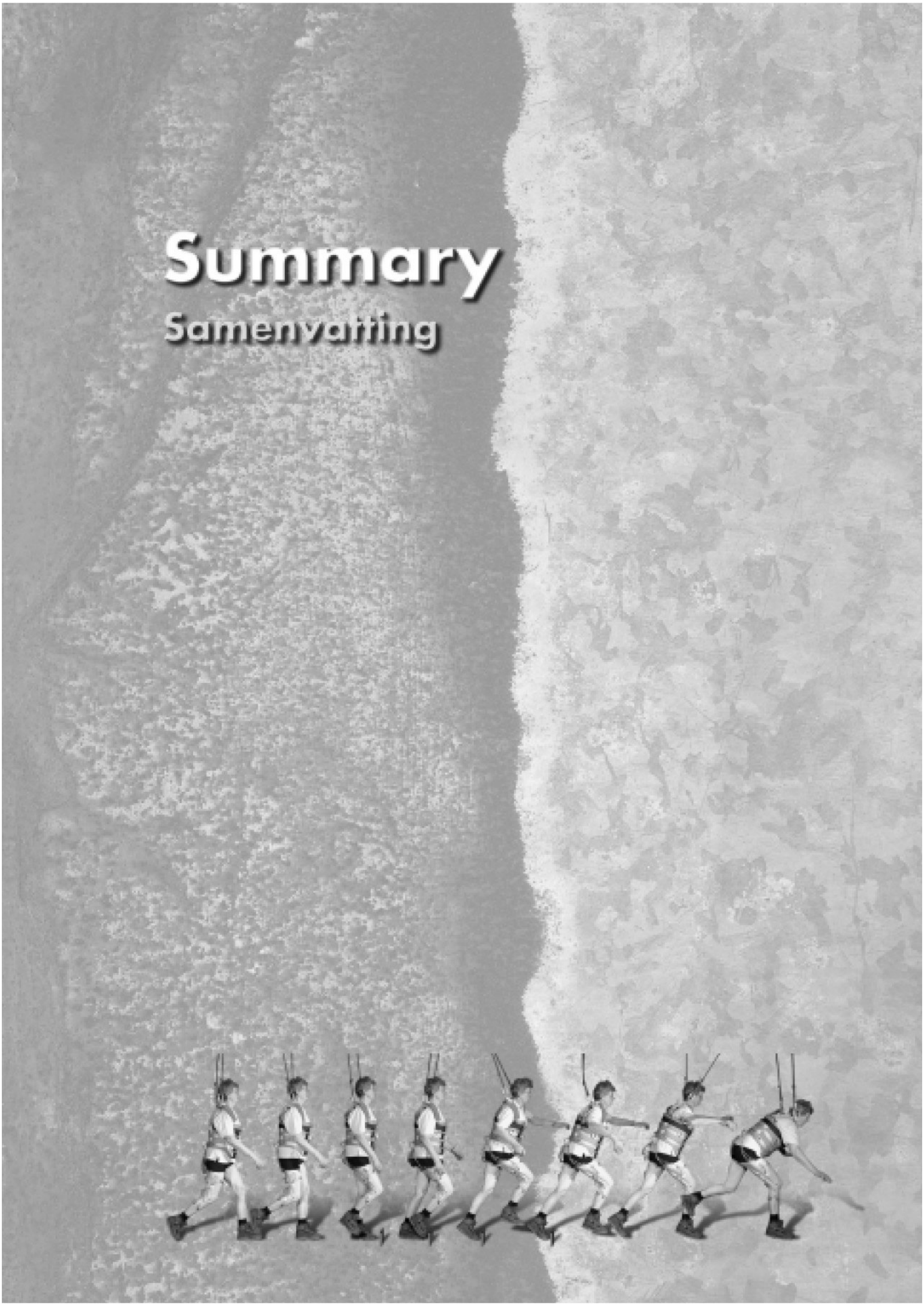
Summary

Samenvatting



Recovery from a trip in young and older adults

Falls and fall-related injuries are the cause of serious medical and social problems, especially in the growing elderly population. Among the community-dwelling elderly people, 25% of the people over 65 years and 35% of the people over 75 years experience at least one fall per year. About 50% of elderly people who fall, experience multiple falls within one year. Falls can result in injuries such as hip and distal forearm fractures, head traumas, and musculoskeletal injuries. Furthermore, the consequences of falls can lead to restricted activity, immobility, adverse psychological effects, admission in a nursing home, or even death. Tripping is one of the main causes of falls in older people. In the general introduction, Chapter 1, it was questioned whether older people fall more often than young people because they trip more often or because they are less able to regain balance after a trip. The probability of tripping might be related to age-related changes in the walking pattern and the fitness of the individual, but the question remains whether older people trip more often than young people. Yet, the probability of recovering successfully from a trip has clearly been indicated in the literature on tripping to be lower in elderly subjects than in young adults.

The main purpose of recovery after tripping is to reduce the angular momentum, which the body has obtained from impact with the obstacle. An inadequate reaction will lead to a fall (Chapter 1). Two recovery strategies after tripping can be discerned; which one of these two occurs depends on the time of trip initiation in the swing phase. An elevating strategy is usually observed after a perturbation in early swing and consists of an elevation of the obstructed (ipsilateral) swing limb to overtake the obstacle. A lowering strategy is usually seen after perturbation in late swing and consists of an immediate placement of the obstructed foot, followed by a next step in order to overtake the obstacle. The limb that is positioned forward after the trip is coined the recovery limb. Angular momentum can be restrained by proper placement of the recovery limb, anteriorly of the body, but the contralateral support limb may also play an important role by providing a powerful push-off. Generating sufficiently rapid and powerful push-off reactions by the support limb could be unfeasible for elderly people, since lower extremity strength and the rate of force generation are known to decline with age. The aim of the present thesis was to obtain insight into the requirements for a successful recovery reaction after tripping - in particular the mechanics and control of that reaction of the support limb - and to understand why older people sometimes fail to meet these requirements.

These insights may help to identify causes for inadequate reactions and falls, and to find targets for intervention.

First, an experimental setup was developed in which subjects could be tripped repeatedly during over-ground walking. As the subjects had to be informed about the purpose of the study for ethical reasons, anticipatory changes in the walking pattern could occur, causing recovery reactions to be different from recovery reactions in real life. Chapter 2 presents a study on possible changes in the walking pattern of young subjects after forewarning of a possible trip. Kinematics and kinetics of normal walking and forewarned walking were compared. Only small increases were found in step width and foot clearance, which were attributed to increased dorsiflexion in the ankle. The observed changes were small and therefore not expected to alter the probability of tripping or the recovery reactions after tripping in an experimental setup. The fact that kinematic patterns were minimally affected indicated that the underlying net joint moment patterns were minimally changed. Theoretically, this could coincide with increased co-contraction of antagonists, leading to more stiffness and damping in case of perturbations. Consequently, the recovery reactions could be affected. Furthermore, the aforementioned study focused on young subjects only, whereas tripping experiments could and would be performed on older subjects. Therefore, a subsequent study on the effect of forewarning on muscle activity patterns in both young and older subjects was performed and presented in Chapter 3. This study revealed that the changes in the walking pattern of older subjects after forewarning was similar to the changes assessed in young subjects. Overall, a clear tendency towards increased activity in antagonistic muscles was found after forewarning in young as well as in older subjects. Although this tendency was strong enough to result in significant effects in some leg muscles, the increased muscle activity was only marginal when compared to the magnitude of muscle activity during tripping responses. Based on the results of Chapters 2 and 3 it was concluded that valid experimentation with respect to tripping reactions is possible.

Secondly, the role of the support limb in recovery from a trip was investigated in young subjects. Chapter 4 revealed that push-off by the support limb contributes to recovery (a) by providing time and clearance for proper positioning of the recovery limb (linear acceleration), and (b) by restraining the angular momentum of the body during push-off (angular deceleration). Quantification of angular momentum was based on calculation of the external moment (M_{ext}), which equals the rate of change in the angular momentum of the body. All subjects were able to provide time and clearance for proper positioning of the recovery limb during push-off by the support limb. Furthermore, although not all subjects were able to reduce the angular momentum completely to zero during push-off, almost all showed a clear attempt to restrain the angular momentum. Chapter 5 dealt with the question of how this push-off reaction is achieved by the support limb. Rapid responses (60-80 ms) in triceps surae and hamstring muscles of the support limb generated a large ankle plantar flexion moment, a knee flexion moment and a hip extension moment. For the hip and knee, these moments have a sign opposite to those during normal walking. This indicates a fast relaxation of hip flexors and knee extensors and rapid build up of force in their antagonists. Concluding from the results of Chapters 4 and 5, the support limb plays an important role in recovery from a trip. Of course, this does not mean that the recovery limb is not important in tripping, since that limb has to be positioned adequately, and should further establish balance after landing. However, the more the angular momentum is reduced by the support limb, the less remains to be accomplished by the recovery limb. In addition, the more time and clearance is provided by the support limb to swing the recovery limb forward, the more appropriate the recovery limb can be positioned (i.e. anteriorly of the body), and the more effectively that limb can reduce the remaining angular momentum.

Thirdly, limitations of older subjects in the mechanics and control of the support limb recovery reactions after tripping were investigated. Many studies have shown that muscle strength as well as the rate of force development is lower in older subjects as compared to young subjects. It was therefore hypothesized that older subjects are less able to generate sufficiently rapid and powerful moments than those observed in the young subjects. In a follow-up study, described in Chapter 6, this hypothesis was tested. It was found that older subjects used similar strategies to regain balance as young adults. Onsets of moment generation did not show substantial differences. However, older fallers showed insufficient reduction of the angular momentum during push-off and less proper placement of the recovery limb. This was due to a lower rate of change of moment generation in all support limb joints and a lower peak ankle moment (all normalized for body mass). Improvement over trials was ascribed to a better positioning of the recovery limb, as no clear advancements were seen in the moment generation by the support limb. Changes in muscle properties with increasing age are known to cause a decline in muscle force and rate of force development and thus could explain the slower moment development during balance recovery in the older subjects. However, changes in motor control could also contribute to the inadequacy of recovery reactions. In Chapter 7, it was shown by means of electromyography (EMG) that the timing and sequencing of support leg muscle responses after tripping is robust to the effects of aging, whereas the magnitude and rise time of the muscle activation decline with age. Response patterns were qualitatively similar between young and older subjects. An increased EMG response latency was found only for the soleus muscle of older subjects, and although statistically significant, this increase was negligible in magnitude (11 ms). Quantitatively, the time needed for muscle response activation was found to be longer in the older subjects than in the young subjects. In addition to changes in muscle properties, these changes in the generation of control signals reduce the rate of force generation, which can hamper the recovery reaction and lead to a fall.

In Chapter 8, the epilogue, the main findings and practical implications of this thesis were discussed. The results of the studies described in this thesis have contributed to a better understanding of the mechanics and control of recovery after tripping in both young and older subjects. The ability to rapidly generate large muscle moments is important for successful recovery after tripping, and has been shown to decline in older fallers. Muscle strength, as well as the rate of activation seem to be limiting factors in the recovery of older people. Strength training may be indicated in older people to reduce the risk of falling after a trip. Training could be targeted at specific muscles groups that, according to this thesis, play an important role in recovery after tripping (i.e. hip extensors and plantarflexors for push-off and possibly knee extensors for landing). As it was found that older fallers improved their recovery success over trials (Chapter 6), more specific training, such as applying repeated perturbations simulating tripping in a safe environment, may also be beneficial. Further research is required to investigate the effects of different interventions on the ability to recover from a trip.

Balansherstel na struikelen bij jongeren en ouderen

Vallen en valgerelateerde letsels kunnen ernstige medische en sociale problemen veroorzaken, met name bij ouderen. Ongeveer 25% van alle mensen boven de 65 jaar valt zeker eens per jaar. Van de mensen boven de 75 jaar valt zelfs 35% eens per jaar. Ongeveer de helft van de ouderen die een keer gevallen zijn, valt meer dan eens per jaar. Valongevallen kunnen resulteren in letsels zoals heup- en polsfracturen, hoofdwonden en aandoeningen aan het spier-skeletsysteem. Bovendien kan vallen bij ouderen beperkingen in activiteiten en mobiliteit, psychologische problemen, noodzakelijke opname in een verpleegtehuis en zelfs sterfte tot gevolg hebben. Struikelen is één van de belangrijkste oorzaken van vallen bij ouderen. In de algemene inleiding (hoofdstuk 1) van dit proefschrift wordt afgevraagd of ouderen vaker struikelen dan jongeren, of dat zij minder goed in staat zijn om hun balans te herstellen wanneer zij struikelen. De kans op struikelen is afhankelijk van de veranderingen met leeftijd in het looppatroon en in fysieke gesteldheid, maar het is niet bewezen dat ouderen vaker struikelen dan jongeren. Uit de literatuur blijkt echter wel dat de kans op vallen na struikelen bij ouderen beduidend hoger is dan bij jongeren.

Voor succesvol balansherstel na struikelen is het noodzakelijk om het impulsiemoment (de hoeveelheid rotatoire beweging dat het lichaam krijgt na botsing met het obstakel) voldoende af te remmen. Een inadequate reactie zal resulteren in een val (hoofdstuk 1). In de literatuur zijn twee strategieën voor balansherstel beschreven. Welke van de twee plaatsvindt, hangt af van het tijdstip van botsing van het zwaaibeen met het obstakel. Wanneer het zwaaibeen vroeg in de zwaaifase tegen een obstakel botst, zal dit been direct over het obstakel heen getild worden. Deze strategie heet de ‘optil-strategie’. Wanneer het zwaaibeen laat in de zwaaifase tegen een obstakel botst, zal dit been onmiddellijk neergezet worden, gevolgd door een stap met het andere been over het obstakel. Deze strategie wordt de ‘plaatsings-strategie’ genoemd. Het been dat naar voren wordt geplaatst, heet het herstelbeen. Dit herstelbeen kan het impulsiemoment afremmen, indien het vóór het lichaam gepositioneerd wordt. Daarnaast kan ook het standbeen (het been dat op de grond staat op het tijdstip van botsing met het obstakel), een belangrijke bijdrage leveren aan succesvol balansherstel door een krachtige afzet te genereren. Het genereren van een snelle en krachtige afzet door het standbeen zou een probleem kunnen zijn voor ouderen, aangezien bekend is dat spierkracht en de snelheid van krachtsopbouw afneemt met leeftijd. Het doel van de studies in dit proefschrift was om inzicht te krijgen in de vereisten voor een succesvol balansherstel na struikelen (in het bijzonder de mechanica en sturing van de reacties in het standbeen) en om te begrijpen waarom ouderen soms niet aan deze vereisten kunnen voldoen. De verkregen inzichten in de oorzaken voor vallen kunnen worden gebruikt in interventies voor valpreventie.

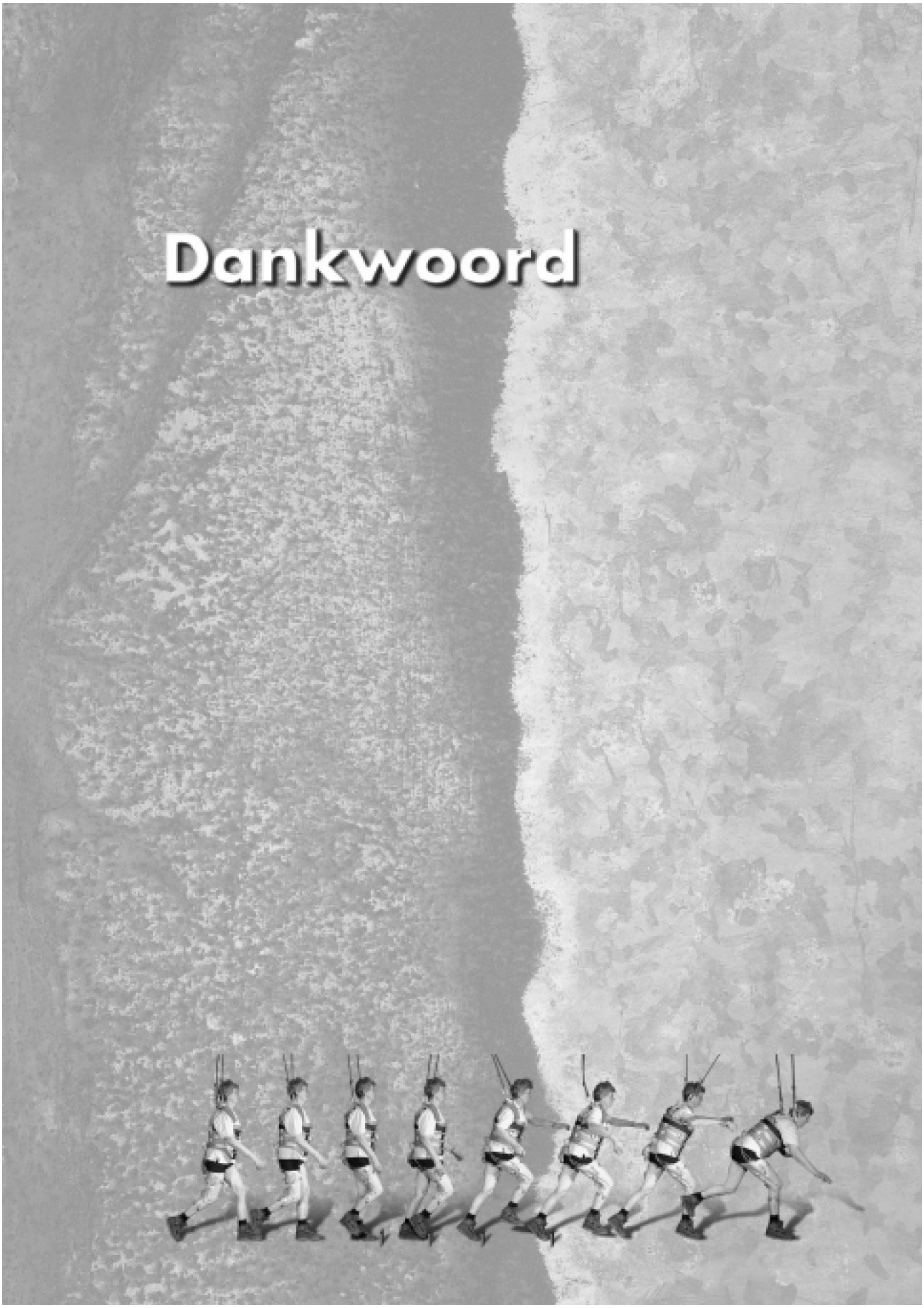
Allereerst werd een experimentele opstelling ontwikkeld, waarmee proefpersonen herhaaldelijk tot struikelen konden worden gebracht, terwijl zij over een vaste ondergrond liepen. De proefpersonen werden om ethische redenen op de hoogte gebracht van het verloop van het onderzoek. Hierdoor wisten proefpersonen dat zij tot struikelen konden worden gebracht en zouden zij hun looppatroon kunnen aanpassen. Door aanpassingen in het looppatroon zouden de herstelreacties in een experimentele omgeving kunnen afwijken van reacties in het dagelijkse leven, wat nadelig zou kunnen zijn voor de struikelexperimenten. In hoofdstuk 2 is een onderzoek beschreven naar de veranderingen in het looppatroon van jong volwassen proefpersonen na waarschuwing voor mogelijk struikelen. De kinematica en dynamica van normaal lopen en gewaarschuwd lopen werden met elkaar vergeleken. Er werden alleen kleine verschillen gevonden in stapbreedte en teenhoogte tijdens de zwaai. De teen werd tijdens de zwaai hoger opgetild door een versterkte dorsaalflexie van de enkel. De waargenomen veranderingen waren echter zeer klein en zullen naar verwachting niet leiden tot veranderingen in de kans op struikelen of in herstelreacties na struikelen in het experiment. Het feit dat de kinematische patronen nauwelijks veranderden na waarschuwing voor mogelijk struikelen impliceert dat de netto momenten in de gewrichten ook nauwelijks veranderd waren. Theoretisch kan dit niettemin samengaan met een verhoogde spierspanning van zowel strek- als buigspieren rondom de gewrichten. Zo'n verhoogde spierspanning in antagonisten kan leiden tot een verhoogde stijfheid en demping in geval van een verstoring, waardoor de herstelreacties mogelijk beïnvloed kunnen worden. Het onderzoek in hoofdstuk 2 betrof bovendien alleen jongvolwassen proefpersonen, terwijl struikelexperimenten ook bij ouderen worden verricht. Daarom is een vervolgstudie uitgevoerd naar de effecten van waarschuwing op de spieractiviteit van zowel jongeren als ouderen. Deze studie is beschreven in hoofdstuk 3. De resultaten lieten zien dat waarschuwen voor een mogelijke struikel dezelfde effecten had op de oudere als op de jongere proefpersonen. Bij zowel jongeren als ouderen werd een licht verhoogde activiteit van de beenspieren waargenomen. Hoewel deze tendens systematisch genoeg was om in enkele spieren te resulteren in significante effecten, waren deze toenames in beenspieractiviteit slechts marginaal in vergelijking tot de zeer sterke activiteit na struikelen. Uit de resultaten van hoofdstuk 2 en 3 werd dus geconcludeerd dat valide onderzoek naar struikelreacties mogelijk is in de ontwikkelde experimentele opstelling.

Vervolgens werd bij jong volwassen proefpersonen de bijdrage van het standbeen aan balansherstel na struikelen onderzocht. Hoofdstuk 4 laat zien dat de afzet met het standbeen bijdraagt aan balansherstel door enerzijds te zorgen voor voldoende tijd en ruimte om het herstelbeen adequaat te positioneren (lineaire versnelling) en anderzijds door gedurende de afzet het verkregen impulsiemoment van het lichaam af te remmen (rotatoire vertraging). Alle proefpersonen waren in staat om door de afzet met hun standbeen voldoende tijd en ruimte te winnen om het herstelbeen vóór het lichaam te plaatsen. Bovendien waren bijna alle proefpersonen in staat om gedurende de afzet het verkregen impulsiemoment te verminderen, hoewel niet iedereen het impulsiemoment volledig tot nul kon reduceren. De vraag hoe het standbeen tot de benodigde afzetreactie komt, is behandeld in hoofdstuk 5. Snelle reacties (60-80 ms) in de hamstrings en kuitspieren van het standbeen leiden tot een heup-extensiemoment, een knie-flexiemoment en een groot plantairflexiemoment in de enkel. Voor de heup en knie zijn deze momenten tegengesteld van teken vergeleken met normaal lopen. Dit veronderstelt een snelle inactivatie van de heup-flexoren en knie-extensoren en een snelle krachtsopbouw in hun antagonisten. Het standbeen blijkt, op basis van de resultaten van hoofdstuk 4 en 5, een belangrijke bijdrage te leveren aan balansherstel na struikelen. Dit betekent niet dat het herstelbeen onbelangrijk is, aangezien dat been adequaat gepositioneerd moet worden om verder balansherstel te kunnen bewerkstelligen. Bedenk echter, dat hoe meer het standbeen bijdraagt aan het reduceren van het impulsiemoment, hoe minder het herstelbeen verder hoeft te realiseren. Bovendien, naarmate het standbeen tijdens de afzet meer tijd en ruimte wint om het herstelbeen naar voren te zwaaien, kan het herstelbeen beter gepositioneerd worden en de rest van het impulsiemoment reduceren.

Ten slotte werden de beperkingen van ouderen in de mechanica en sturing van herstelreacties na struikelen onderzocht. Eerdere studies hebben aangetoond dat ouderen minder spierkracht hebben en trager zijn in krachtsopbouw dan jongeren. Ouderen zouden daarom mogelijk minder goed in staat zijn in het snel genereren van grote momenten, zoals waargenomen tijdens het herstel na struikelen bij jongeren. In de studie, beschreven in hoofdstuk 6, werd deze hypothese getoetst. Oudere proefpersonen bleken dezelfde strategieën voor balansherstel na struikelen te laten zien als jongeren. Ook de reactietijden waren niet verschillend tussen jongeren en ouderen. Oudere vallers bleken echter onvoldoende in staat om het impulsiemoment af te remmen tijdens de afzet met het standbeen en om het herstelbeen vóór het lichaam te positioneren. Deze tekortkomingen waren toe te schrijven aan een tragere opbouw van de momenten rondom alle gewrichten van het standbeen en aan een lager piekmoment in de enkel (alle genormaliseerd voor lichaamsgewicht). Gedurende het experiment was over achtereenvolgende struikelreacties verbetering te zien in het balansherstel van de oudere vallers. Deze verbetering werd toegeschreven aan een betere positionering van het herstelbeen, omdat geen duidelijke verschillen zichtbaar waren in de gegenereerde gewrichtsmomenten van het standbeen. Het is bekend dat spierkracht en snelheid van krachtsopbouw kunnen verminderen met toenemende leeftijd door veranderingen in spiereigenschappen. Hierdoor kan de tragere opbouw van momenten tijdens het balansherstel na struikelen van ouderen verklaard worden. Naast veranderingen in spiereigenschappen kunnen bij ouderen ook veranderingen in de bewegingssturing de oorzaak zijn van inadequate herstelreacties. In hoofdstuk 7 werd aangetoond, op basis van registraties van spieractiviteit, dat de reactietijden en volgorde van aansturing van de spieren van het standbeen niet veranderen met leeftijd. De spierresponsie-patronen waren namelijk kwalitatief hetzelfde voor jongeren en ouderen; een marginaal toegenomen reactietijd (met 11 ms) werd slechts in één van de kuitspieren (m. soleus) gevonden. Kwantitatief werden wel leeftijdsverschillen gevonden: de grootte en snelheid in toename van de spieractiviteit waren aanzienlijk lager bij ouderen. Deze veranderingen in aansturing kunnen, naast de veranderingen in spiereigenschappen, een negatief effect hebben op de snelheid van krachtsopbouw. Hierdoor kunnen de herstelreacties van ouderen inadequaat zijn, hetgeen uiteindelijk kan leiden tot een val.

In hoofdstuk 8, de epiloog, worden de bevindingen en praktische implicaties uit dit proefschrift besproken. De resultaten van de beschreven studies hebben bijgedragen aan een beter inzicht in de mechanica en sturing van balansherstel na struikelen bij zowel jongeren als ouderen. Voor een adequaat balansherstel blijkt het snel genereren van grote gewrichtsmomenten belangrijk. Naast afgenomen spierkracht is vooral ook de snelheid van krachtsopbouw een beperkende factor voor balansherstel bij ouderen. Op basis van deze resultaten wordt krachttraining bij ouderen aanbevolen, teneinde de kans op vallen en valletsels te reduceren. Training zal met name gericht moeten zijn op de spiergroepen die een belangrijke rol spelen bij balansherstel na struikelen. Dit zijn de heup-extensoren en plantairflexoren voor de afzet en mogelijk de knie-extensoren voor het neerkomen. Verder bleken oudere vallers in staat om hun herstelreacties te verbeteren gedurende het experiment. Training op een meer functioneel niveau, bijvoorbeeld door herhaaldelijk te struikelen in een veilige omgeving, zou daarom ook effectief kunnen zijn. Verder onderzoek zal moeten uitwijzen wat bij ouderen de effecten zijn van (verschillende) trainingen op spiereigenschappen en op balansherstel na struikelen.

Dankwoord



lo·pen (onov.ww.)

zich te voet voortbewegen, gaan

strui·ke·len (onov.ww.)

het evenwicht verliezen of vallen door met de voet ergens achter te blijven haken; een misstap

val·len (onov.ww.)

met een zekere snelheid door de lucht omlaaggaan ten gevolge van de werking van de zwaartekracht; plotseling en onvrijwillig op de grond terechtkomen

dan·ken (ov.ww.)

zijn erkentelijkheid tot uitdrukking brengen tegenover iemand

Dankwoord

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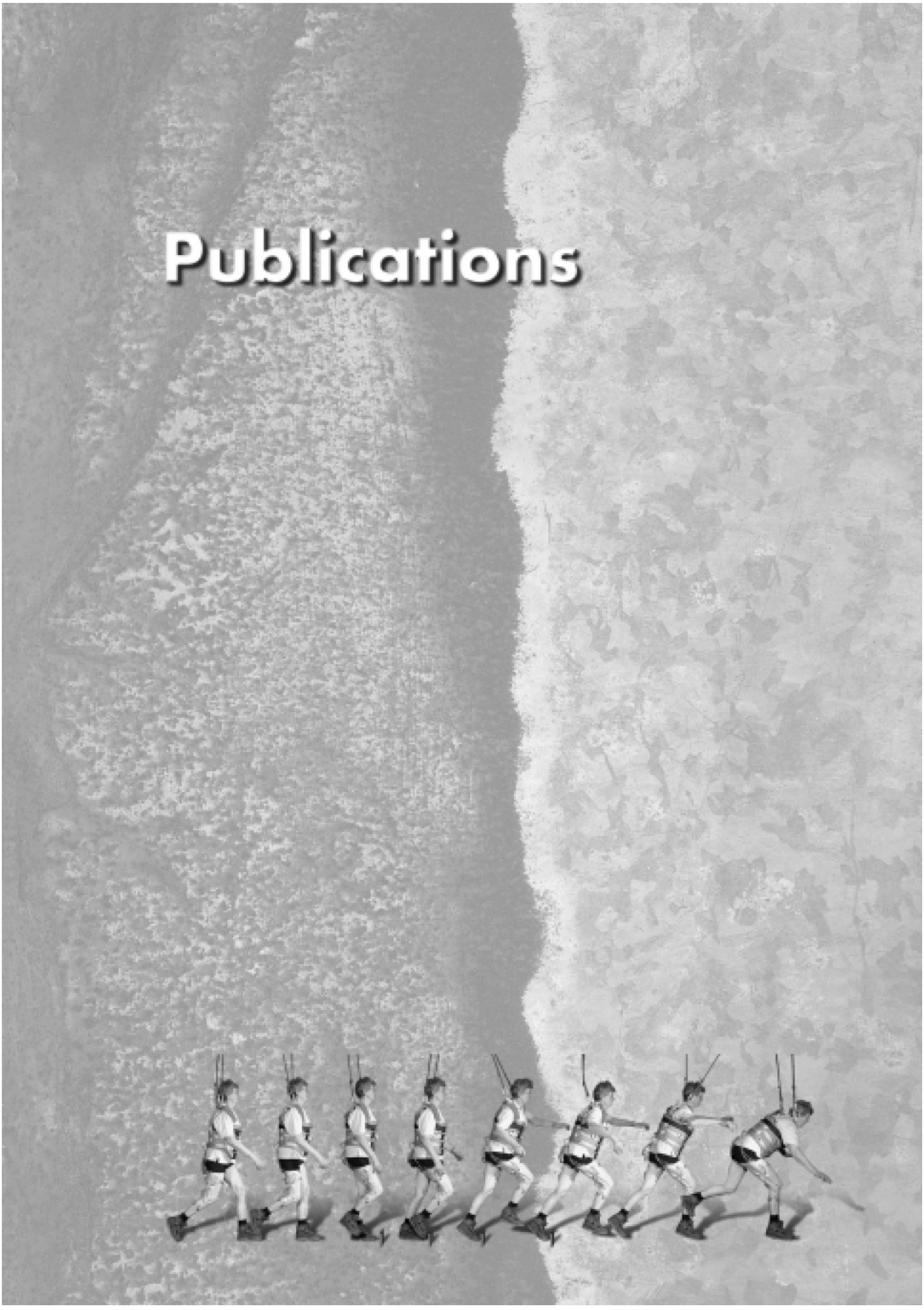

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